Patient Lifter Mechanisms: Forces, Trajectories, and Adjustments for Customer and Mechanism Variations

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ABSTRACT

New Zealand's population is ageing; it is predicted, by 2051, 25 percent of the population will be aged 65 or over. This increase in elderly population is expected to put a larger demand on nursing staff, increasing the amount of patient handling required. Patient handling is defined as the moving and transporting of people with mobility issues. Musculoskeletal injuries during patient handling tasks are currently the most common injury affecting nursing staff in New Zealand. While it is largely documented that these injuries can be decreased by the implementation of lifting programs and installation of lifting equipment, some staff remain resistant to their use due to the time taken and complexity of existing devices.

To address this need, a fully mechanical patient lifter has been developed. It is anticipated that the simplicity and compactness of a fully mechanical lifter would be advantageous, allowing the device to be effective in a large range of situations. The lifter's key benefit is predicted to be the reduction in time taken to complete a lift. The absence of electrical componentry removes the need for regular electrical certification, ongoing battery charging and maintenance, and results in a reduced footprint, allowing easier storage and manoeuvring of the device. Using carer-driven lifting force allows the speed of the lift to be tailored to each patient, reducing the time required to lift more able patients. Due to the simplicity and intuitiveness of a fully mechanical lifter, the training time would be reduced with less refresher and retraining sessions required.

The objective of this study was to assess the performance of a fully mechanical patient lifter and provide recommendations on possible developments. Key performance indicators were developed into an evaluation matrix to evaluate developed mechanisms against transfer aids already available to the public. A key focus of this development was the assessment of handle force required to lift a patient. Theory regarding patient and mechanism forces has been developed and validated using motion capture and a load cell to validate this theory.

The impact of anthropometric variation in patients was also assessed. It was found that, while the main diversity factors were height and weight, a patient's centre of mass location was also important. To this end, it was found the most suitable configuration for a fully mechanical patient lifter was a double pivot mechanism. This mechanism gave the ability to adjust the gradient of the patient's lifted centre of mass (LCM) trajectory, to decrease handle forces, while allowing the total change in height of the LCM to be greater, providing greater patient comfort. Mechanism characteristics, including kneepad, chest pad and pivot placement, type of mechanism, chest pad, and mechanism dimensions, were also assessed to ensure that a suitable balance was achieved between handle forces and patient comfort.

Contents

TABLE OF FIGURES

NOMENCLATURE

Note: *Forces are taken as positive in the right (x) and upwards (y) directions. Angles are taken as positive in the anticlockwise direction from the positive vertical direction unless otherwise stated. Moments are taken as positive when acting in the clockwise direction. Where necessary, the origin of all measurement point on the footplate at the tip of the patients toe unless otherwise stated*

a	ALP to LCM: Horizontal distance from the ALP to the LCM (m)
b	ALP to Hip: Distance from the ALP to the Hip joint (m)
c	Non-ALP Chord Segment: Used in fitting a single pivot path to the zero force trajectory, value of ALP pivot diameter less ALP chord length height (m)
d	Thigh Length: Length of thigh (m)
ALP	Alternate Lift Point: Point around which the torso of the patient rotates (m)
ALPD	ALP Distance: Perpendicular distance from torso to ALP position (m)
ALPN	ALP Normal Line: Normal line of ALP, perpendicular to the trajectory of the ALP path
$ALPx_{PB}$	ALP Chord Midpoint (x): Horizontal component of the midpoint of the ALP chord segment (m)
$ALPy_{PB}$	ALP Chord Midpoint (y): Vertical component of the midpoint of the ALP chord segment (m)
ALPR	ALP Radius: Distance of ALP from hip joint, in ALP Angle direction (m)
ALPx	ALP X Position: Horizontal component of ALP position (m)
ALPy	ALP Y Position: Vertical component of ALP position (m)
CFx	Horizontal Chest Pad Force: Force applied by the patient to the mechanism at the ALP in the horizontal direction (N)
CFy	Vertical Chest Pad Force: Force applied by the patient to the mechanism at the ALP in the vertical direction (N)
CH_{ALP}	ALP Chord Height: Chord height of the arc of the ALP trajectory (m)
CL_{ALP}	ALP Chord Length: Chord length of the arc of the ALP trajectory (m)
CMH	Handle Centre of Mass: Calculated centre of mass of lifter handle and chest pad (m)
COM_T	Thigh Centre of Mass: shown as a percentage of the distance along the thigh from the knee
DSR	Design Specification Rating: Result of mechanism assessment against design specifications and matrix
FEI	Equilibrating Input Force: Force along the ALP trajectory necessary to hold the patient at the current position (N)
FC	Resultant Chest Pad Force: Combined vertical and horizontal components of chest pad force (N)
FK	Kneepad Force: Force acting on the patient from the kneepads of the lifter (N)

FS	Shank Force: Internal force acting on the patient's shank (N)
H	Patient Height: Height of the Patient, when standing (mm)
HF	Handle Force: Force applied to handle to lift patient (N)
Hip_x	Hip Point (x): Horizontal position of hip joint (m)
Hip_y	Hip Point (y): Vertical position of hip joint (m)
HR	Handle Radius: Perpendicular distance from IPP to carer handle (m)
HW	Handle Weight: Weight force due to handle geometry (N)
HWx	Handle Weight Distance: Horizontal distance between Initial Pivot Point and Handle Centre of Mass (m)
IPPx	Initial Pivot Point (x): Horizontal position at which the Primary Pivot Arm is secured (m)
IPPy	Initial Pivot Point (y): Vertical position at which the Primary Pivot Arm is secured (m)
LCM	Lifted Centre of Mass: Centre of Mass position for the lifted section of the body, namely the Head, Neck, Torso, Arms and half of the thigh weight (m)
L_T	Torso Length: Length of torso, from hip joint to shoulder joint
Px	Horizontal Pivot Force: Resultant force in horizontal direction at pivot point of single pivot (N)
Py	Vertical Pivot Force: Resultant force in vertical direction at pivot point of single pivot (N)
R1	Primary Pivot Arm Length: Length of the pivot arm between the IPP and ALP (m)
R2	Secondary Pivot Arm Length: Length of pivot arm between the ALP and Chest pad (m)
Rx	Foot Reaction Force (x): Horizontal force applied by the patient to the footplate of the lifter, exclusive of shin and foot weight (N)
Ry	Foot Reaction Force (x): Vertical force applied by the patient to the footplate of the lifter, exclusive of shin and foot weight (N)
SPPx	Secondary Pivot Point (x): Horizontal position at which the Secondary Pivot Arm is secured (m)
SPPy	Secondary Pivot Point (y): Vertical position at which the Secondary Pivot Arm is secured (m)
TH	Thigh Weight at Hip: Force due to thigh weight (T) acting at the hip (N)
T	Thigh Weight: Weight of the thighs, lower leg and foot; applied as a point force at the COM_T (N)
TA	Axial Thigh Force: Thigh force along axis of the thigh (N)
TK	Knee-Thigh Force: Force due to T acting at the knee, applied in the vertical direction (N)
TR	Radial Thigh Force: Component of the axial thigh force acting in the direction perpendicular to the ALP line (N)
VLP	Virtual Load Point: Point at which R1 and Hip to Torso line intersect (m)

VLPx	VLP X Position: Horizontal component of the VLP position (m)
VLPy	VLP Y Position: Vertical component of the VLP position (m)
W	Weight Force: Self-Weight force of the lifted weight of the patient (N)
WT	Thigh Weight at Knee: Force due to T acting at the knee, reaction force to TK (N)
α	Torso Angle: Angle of the torso (deg)
β	Thigh Force: The angle of the thigh anticlockwise from the positive horizontal axis (deg)
γ	Alternate Lift Point Angle: The angle from the torso angle to the ALP (deg)
δ	Resultant Chest Pad Force Angle: Angle of the resultant chest pad force (deg)
ε	ALP Path Normal Line Angle: Angle of the normal line to the ALP path (deg)
ζ	ALP Coefficient: Location of contact point of chest pad with torso, as a decimal of total torso length
η	ALP Chord Angle: Angle of ALP trajectory chord from vertical (deg)
θ	Resultant Angle: Difference in angle between Resultant Chest Pad Force Angle (δ) and ALP Path Normal Angle (ε) (deg)
λ	Shank Angle: Angle of lower leg anticlockwise from the negative vertical direction (deg)
μ	R2 Angle: Angle of the Secondary Pivot Arm anticlockwise from the positive horizontal direction (deg)

1 INTRODUCTION

1.1 IMPLICATIONS OF AN AGEING POPULATION

Expected population ageing is forecast for New Zealand, with the median age expected to rise from 35 in 2003 to 46 by 2051 (Cornwall & Davey, 2004) . In 2007, 12.3 percent of the New Zealand population was aged 65 or over. It has been projected that this could rise to 25 percent of the population by 2051 (Statistics New Zealand, 2007). The 65 and over age group is also a growing consumer group with total spending amounts rising from $14 billion in 2011 to $65 billion in 2051 (Office for Senior Citizens, 2015). These projected increases are expected to put pressure on health providing systems resulting in the encouragement of informal homecare and community-based assistance services. It is expected that the number of older people in informal care situations could rise by 56 percent between 1995 and 2031 (Cornwall & Davey, 2004).

Within this growing age group, it has been found that the most common type of disability or restriction is mobility related with 39 percent of men and 46 percent of women aged 65 or over suffering from mobility restrictions (Cornwall & Davey, 2004). The moving and transporting of people with mobility issues falls under the term “patient handling”. Patient handling can be very difficult to undertake in a considerate manner due to the large loads and unpredictability involved (Accident Compensation Corporation, 2003). The benefits of utilising assistive mobility and lifting aids is well-documented (Borner, 2008; Accident Compensation Corporation, 2012; Li, Wolf, & Evanoff, 2004).

1.2 THE DEFINED NEED

Lifting mechanisms currently available within the patient handling market are summarised in Sections 2.6.1 - 2.6.6. The current solutions for patient handling are mostly large, slow, and expensive or require the patient to be able to stand independently with relatively good balance. It has become apparent that there is a gap within the current patient handling market for a device to transfer patients between seated positions that is inexpensive, intuitive, safe, and simple.

To provide a lifting aid that meets these criteria, it was decided to develop a fully mechanical mechanism. The simplicity of a fully mechanical design would reduce training times, maintenance, and reliability issues. Utilising mechanical advantage and fully understanding the forces present are expected to result in a mechanism that is simple, cost-effective, and reliable.

Prior to this project, many mechanisms were considered to fill this gap. A common issue with these regarded altering the mechanisms to produce reduced handle forces. The setup of a mechanism has been termed “lifter characteristics” and includes possible adjustments to the geometry of the lifter including knee pad, chest pad, mechanism, and pivot placements. These uncertainties limit the functionality, effectiveness, and practicality of the lifter. Knowledge of how, and why, the forces alter is imperative to the project, as being able to limit and adjust the forces would result in a larger range of patients and carers being able to use the device.

To develop an effective mechanism, the forces felt by the patient also need to be assessed and considered. As well as providing insight into the patient comfort during a lift this will also aid in understanding the forces applied by the patient to the mechanism. An understanding of the implications of patient characteristics on these forces is also of importance. Patient characteristics are defined as the size and shape of a patient and includes total weight, total height, waist circumference, body mass index, mobility and health issues, patient’s height and weight distributions, and location of the lifted centre of mass. This will aid in developing an understanding of expected handle forces for a variety of patients and how to reduce this. The scope and methodology are defined in Appendix A.

2 Review of Literature

Several major aspects relating to this topic are discussed in the following Literature Review; namely, the effect of the ageing population, patterns in community-assisted and informal care situations, changes in the healthcare sector, the measuring of carer forces during the use of transfer aids, device requirements, and existing solutions available for patient handling.

2.1 The Ageing Population

In New Zealand, the ageing population is expected to influence the Healthcare sector in a two-fold manner, by causing an increase in demand on geriatric healthcare whilst causing a decrease in the number of healthcare professionals. Whilst the general population median age is rising, it is expected that the median age within the over sixty-five demographic will also rise from 74.2 years in 2007 to 77 years by the year 2051 (Statistics New Zealand, 2007). It is also expected that the median age of the over sixty-five demographic will continue to increase irrespective of the median age increase within the total population (Statistics New Zealand, 2009). It should also be noted that the ageing population is not evenly spread throughout New Zealand, with some smaller District Health Boards, such as South Canterbury and Wairarapa having 15 percent of their population over 65. In these areas, this is expected to rise to 25 percent by 2021 (Ministry of Health, 2002).

An important economic concern for countries with ageing populations is the projected increase in healthcare expenditure. For the over sixty-five demographic, the healthcare expenditure per person is 3.8 times larger than the healthcare expenditure per person below 65 years, this increases to approximately five times the average healthcare expenditure per person for the over seventy-five demographic (Carey, 1999). For 2000/01 the over 85 age group accessed on average nine General Practice visits, \$642 and \$629 of subsided pharmaceuticals for women and men respectively, and ten laboratory tests per year. This is compared to six General Practice visits, \$423 of subsidised pharmaceuticals, and seven laboratory tests per year for the 65-74 demographic (Ministry of Health, 2002). Hospital admission data from 2001 in New Zealand show that the over sixty-five demographic represented 32 percent of medical and surgical admissions to hospitals, excluding maternity admissions. A large percentage of this increased healthcare expenditure can be attributed to the Government funding of nursing home and hospice level care, although the documentation on the actual amount is limited. If this percentage were to be quantified, it would be a useful tool to calculate the savings suitable in-home care and support would generate.

Disability in the elderly is an important factor as it can greatly affect independence and quality of life. Through the 2013 New Zealand Disability Survey it has been found that 59 percent of the over sixty-five demographic identified as disabled (Statistics New Zealand, 2014). This is a much higher percentage than the total population, where disability rates are around 25 percent. The most notable difference in percentages between demographics was in the category of physical impairments, where 7 percent of persons under 45 years were physically disabled compared to 49 percent of persons 65 years or over (Statistics New Zealand, 2014) . Physical disabilities are defined as mobility or agility impairments. These patterns present within the ageing population indicate that the issue of safe patient handling is likely to increase as a concern.

2.2 INFORMAL CARE

Expected increases in healthcare demand, along with an ageing nursing staff, informal care in which a person remains in a private residence supported by community organisations, has become a more popular and vital sector of healthcare. The number of older people in informal care situations is expected to rise by 56 percent between 1995 and 2031 (Cornwall & Davey, 2004). This is consistent with the majority of European countries (Sorbye, 2009). In 2001, the New Zealand Office for Senior Citizens developed the Positive Ageing Strategy designed to keep older people engaged and able to participate in their local communities (Office for Senior Citizens, 2008). Currently in New Zealand, almost one in ten people are informal carers, with around 60 percent of these carers being women (Ministry of Social Development, 2014). The largest growing populations of carers are older people caring for older people and mid-life carers caring for older people (Ministry of Social Development, 2014). One of the major limitations with the elderly remaining in the community, or “ageing in place”, is the effect of a disability on their standard of living. In 2014, it was found physical impairments were the most commonly cited disability impacting on people’s living situations, with the majority of physical impairments in people over sixty-five being mobility related (Statistics New Zealand, 2014).

House ownership is high in this demographic with 76 percent of over sixty-fives owning or partly owning their own home. This percentage decreases to around 50 percent for people aged over 90 (Statistics New Zealand, 2007). Approximately 7 percent of over sixty-fives live in non-private dwellings, with the majority of these living in residential care facilities (Statistics New Zealand, 2007). It is clear from Wiles, et al. (2017) and Hayman, et al. (2012) that “ageing in place” is an ideal valued by the older person, with the over 90 percent of participants not anticipating moving in the near future, but one very inconsistent with the findings of the housing section of the New Zealand Census summary (Statistics New Zealand, 2007). It may have been beneficial to ask participants to outline any reasons they would foresee themselves moving and what community support they would expect to assist them in continuing to live in their private homes. In many cases, it has been found that older people’s health and wellbeing is strongly linked to their social and physical environments (Wiles, et al., 2009).

The 2000 National Home and Hospice Care Survey found that the percentage of home health care patients reliant on mobility aids greatly increased with age, from 21 percent of the under 18 age group, to 78 percent of the over 85 age group (Centers for Disease Control and Prevention, 2000). Of the respondents with mobility aids, it is noted that the greatest increase in mobility aid usage was the use of walkers which increased from 36 percent in the 18-44 years age group to 69 percent in the over 85 age group. It can be extrapolated from this that the majority of respondents are still relatively mobile but lack the stability and coordination to stand and walk fully unaided and would require patient handling equipment.

2.3 HEALTHCARE SECTOR

Staffing in the healthcare sector are facing many issues. While confronting rapidly increasing patient numbers, the sector is also suffering high turnover and staff shortage. In the United States of America, it is expected that the Healthcare Support Occupations of nursing, aides, and orderlies will increase by 23.6 percent between 2016 to 2026. During this time, the healthcare and social assistance sector are projected to require nearly four million new employees (Bureau of Labor Statistics, 2018). It is estimated that average yearly turnover of nursing staff in New Zealand hospitals is around 40 percent (Borner, 2008). The conclusion can be drawn that high staff turnover impacts the knowledge level of industry-specific practices such as the handling of patients. These turnover rates result in an increase in inexperienced staff, decreasing productivity, job satisfaction, and patient care quality (North, et al., 2006). It is expected the increase in demand for healthcare and public services from the ageing population will require changes in policy and restructuring to ensure fiscal sustainability (Buckle & Creedy, 2014). It has been found that this situation is not limited solely to New Zealand, with many countries facing the same issues (Beard & Bloom, 2015).

The percentage of the over sixty-five demographic in the workforce is increasing, with the most common occupation for women over sixty-five in the nursing support and personal care sector (Statistics New Zealand, 2007). These findings are supported by overseas research, which show nursing staff are predominately female, with women making up 93.4 percent of nursing staff in America during 2008 (Vaughan, Driver, Hall, & Race, 2014). With an ageing and predominantly female workforce, manual patient handling is a growing risk to the health of nursing staff.

The New Zealand recommended best practice standards have been taken from guidelines previously used in the United Kingdom. These guidelines recommend that maximum lift weights vary depending on their location respective to the carer, with a maximum of 16 kilograms for women and 25 kilograms for men (Accident Compensation Corporation, 2003). It can be seen from this, that most patient handling operations would have instances where these weights are easily exceeded. Collins, Nelson, & Sublet, (2006) recommend that residential care facilities should provide one full-body machine for every eight to ten non-weight bearing patients and one standing lifter for every eight to ten limited weight bearing patients. It is widely recognised these means of patient lifting are time consuming, bulky, and require specialist training (Schoenfisch, Myers, Pompeii, & Lipscomb, 2011). It is anticipated that a reliable, quick, and intuitive lifting device would be beneficial to an ageing and inexperienced workforce.

2.4 CARER INPUT FORCES

Handling injuries are very common during patient manoeuvring due to the high forces, patient unpredictability, and improper practice techniques. Over $30 million in ACC claims distributed in 1999 were due solely to patient handling injuries; the most common of these patient-handling injuries being musculoskeletal (Accident Compensation Corporation, 2003). In the United States of America Healthcare sector, the incident rate for musculoskeletal disorders during 2005 was 82.3 cases per 10,000 workers, with the majority of these being caused by manual patient handling (National Occupational Research Agency, 2009). Whilst New Zealand is facing challenges to its economic sustainability from an ageing population, challenges also arise from the increase health expenditure due to obesity (Lal, Moodie, Ashton, Siahpush, & Swinburn, 2012). Obesity is an increasing issue in New Zealand with 30 percent of New Zealand adults in 2011 classified as obese, with 38 percent of 65-74 year olds classified as obese (Ministry of Health, 2015). The Accident Compensation Corporation provide a working definition of bariatric as "someone who weighs 150kg or more, has a BMI of 40 or more or who has large physical dimensions, a lack of mobility or other conditions that make moving and handling difficult" (Accident Compensation Corporation, 2012, p. 388).

It has been widely accepted that manual patient lifting, without the use of transfer and lift equipment, is dangerous for the workforce and can lead to long term musculoskeletal injuries for carers (Borner, 2008; Alamgir, et al., 2009; Accident Compensation Corporation, 2003). Studies have researched the impact introduction of patient lifters has had to injuries and the relevant costs. Whilst it is found that there is an decrease in patient and carer injuries, it is difficult to determine the impact as this often occurs in tandem with the introduction of new patient handling practices (Schoenfisch, Lipscomb, Pompeii, Myers, & Dement, 2013). It should be noted limitations arise from carers unwilling to use lifters and the underreporting of injuries (Li, Wolf, & Evanoff, 2004).

While extensive research has been completed on the comparison of carer forces when using overhead and floor lifts, limited documentation exists on how to limit the forces during the lifting procedure. This is most likely due to the majority of devices using battery operation to complete the patient lift. Carer input forces have been measured with respect to the moving of patients utilising floor-based lifts using several different methods. Lachance, et al., (2016) completed transfers over a variety of floor surfaces while carer input forces were measured utilising an accelerometer and tri-axial load cell. For this assessment, only the horizontal carer forces during a straight line manoeuvre of already lifted patients was assessed.

Waymouth (2014) utilised a force balance table to analyse the transferred force of the carer to the ground. Similar techniques have been used in testing of carer forces (Dutta, Holliday, Gorski, Baharvandy, & Fernie, 2012). Van der Woude, Geurts, Winkelman, and Veeger (2003) used strain gauge force transducer applied to handles to measure carer pushing forces. Research has also been completed using a lumbar motion monitor to measure position, velocity and acceleration (Marras, Knapik, & Ferguson, 2009). While the manual lift of a patient was not completed, comparison of forces and techniques of the Pull, Turn and Push activities may be beneficial. Many studies chose to assess carer input through the use of surveys to assess the carers perceived effort and overall satisfaction with the lifting procedures (Borner, 2008; Alamgir, et al., 2009). Ha, Cao, and Khasawneh (2014) used Digital Human Modelling Software to calculate comfort assessment results from Rapid Upper Limb Analysis although this is limited in its application completed requires testing to validate. It is also possible to calculate handle forces through the measurement of accelerations applied to a handle for a four-caster manually manoeuvred vehicle such as a patient lifter (Abraham & Johnson, 2010). It is anticipated utilising a reliable and accurate form of force measurement will be important in assessing and reducing forces within a fully mechanical mechanism.

2.5 Assistive Device Performance

Assessment of the suitability and performance of lifting devices within the market can be completed through a set of developed specifications. From the literature review and market analysis, seven key requirements to assess lifting devices were developed namely, ease of use, carer input, safety, stability, cost, manoeuvrability, and cognitive requirement. These are discussed in Sections 2.5.1 - 2.5.7.

2.5.1 Ease of Use

Required time for a mechanical lift is the largest barrier to their use within care situations (Schoenfisch, Myers, Pompeii, & Lipscomb, 2011; Noble & Sweeney, 2018). The time-related issues can be split into mechanism and facility factors, respectively. Facility factors are mostly irrelevant when assessing performance of a certain device. Facility factors that are not influenced by the assistive device can include staffing issues, facility culture, and device availability. The perceived barriers of device availability and staffing levels account for 78 and 79 percent of reported barriers respectively (Noble & Sweeney, 2018).

Time-related issues due to mechanism factors are key to assessing the performance of a device. Schoenfisch, Myers, Pompeii, & Lipscomb (2011) found that nursing staff felt time pressures were present when locating the device, completing the lift, and during staff device training. Obviously, locating the device is mainly a facility factor but a device with a smaller footprint would be more likely to be easily accessible. Time pressure during the lift arises from the length of time a device-aided lift takes compared to a manual lift. A comparison of time to lift a patient manually and with a floor lift shows that manual lifts with sliding sheets take an average 37.2 seconds while floor lifts taken an average of 287.9 seconds (Alamgir, et al., 2009). Device training was also highlighted as a limitation for using assistive devices. In their research, Noble and Sweeney (2018) state 32 percent of respondents reported they lacked the knowledge to operate the assistive devices within their facilities. Schoenfisch, Myers, Pompeii, and Lipscomb (2011) found it can take staff up to four hours to be trained with refresher relearning required by staff with more complex devices when used infrequently. Ideally, a mechanism should be capable of completing a lift in a time comparable to that of a manual lift.

2.5.2 Carer Input

Often, the high forces present in patient handling often leads to injury such as long term musculoskeletal injuries for carers, as discussed in Section 2.4 above (Borner, 2008; Alamgir, et al., 2009; Accident Compensation Corporation, 2003). Dutta, Holliday, Gorski, Baharvandy, and Fernie (2012) found that a 90 kilogram patient using a mobile hoist on a level, hard surface generated maximum carer forces of 151.3 Newtons when manoeuvred. This is consistent with New Zealand recommended best practice of limiting patient handling activities to below 16 kilograms (Accident Compensation Corporation, 2003). From this, it can be taken that carer input force should be less than 16 kilograms or 157 Newtons.

2.5.3 Safety

Note: Patient, carer and mechanism safety has been broken into ten key points, defined from research of current literature:

No Carer Forward Back Bending throughout Lift: It is well documented that lifting with a bent back causes large musculoskeletal strains and is more likely to cause injury (Andersen, Fallentin, Thorsen, & Holtermann, 2016; Holtermann, Clausen, Aust, Mortensen, & Andersen, 2013). The key to the "16 Kilo Limit" is the majority of lifting is completed between elbow and knuckle height, using correct posture and spinal alignment (Accident Compensation Corporation, 2003).

Load Limited to 16 Kilograms throughout Lift: This is recommended to be limited to 13.5 and 20 kilograms above shoulder height for males and females respectively (Accident Compensation Corporation, 2003). It should be noted that the 16 Kilogram Limit is a recommendation for risk assessment or review of procedure should be completed rather than a legal requirement (Accident Compensation Corporation, 2003).

Locking Mechanisms, Emergency Stop, and Emergency Manual Override: To ensure controlled lifting a locking mechanism should be present. An emergency override must also be present in electrical systems with wheel brakes present (Accident Compensation Corporation, 2003).

No Twisting of Carer Torso throughout Lift: Twisting of the torso increases the chances of injury and causes back strain (Andersen, Fallentin, Thorsen, & Holtermann, 2016). The shoulder and pelvis of the carer should remain in line throughout the lift (Accident Compensation Corporation, 2003).

No Dragging of Patient across Lifting Surface: As the majority of users of assistive devices are elderly, special care needs to be taken, as the patient's skin may be delicate and prone to tearing or bruising (Accident Compensation Corporation, 2003). It has been found that effective mechanical lifts can decrease patient transfer injuries by 62 percent (Garg, 1999).

Force Spread over Large Portion of Patient's Body: The force applied to lift a patient should not be limited to a small, or soft tissue, area (Enos, 2008). High forces over a small area can cause bruising and discomfort. As well as comfort throughout the lift, it should also be ensured that patients maintain a suitable level of dignity (Speser, 2011).

Ability to Transfer To and From a Variety of Surface Heights: Adjustability of the mechanism for a variety of patient heights and weights is important to ensure patient safety and comfort (Enos, 2008). A key benefit for an assistive device is the ability to aid patients exiting a car (Accident Compensation Corporation, 2003).

No Pinch Points, Fully Mechanically and Electrically Safe: It should be ensured there are no trapping or pinch points present within the lifter mechanism that could injure either patient or carer. Devices should also be full mechanical and electrical safety (Accident Compensation Corporation, 2003).

Transportable: The devices should be lightweight when unloaded, and easily transportable (Accident Compensation Corporation, 2003). It should be noted that ease of transport relates to the device when not loaded with a patient, whereas manoeuvrability relates to when a patient is present.

Operation Errors Easily Reversed: Easily reversed equipment operation ensures the device is simple to use, and limits the risk and severity of operator error (Enos, 2008). It is expected this will also include the need for intuitive controls for the lift and transport of patients.

2.5.4 Stability

Cornwall & Davey, (2004) note that mobility disabilities effect 39 percent of men and 46 percent of women aged 65 or over. In New Zealand, it was found approximately 12 percent of total population suffer from mobility impairments. Mobility impairments are defined as having difficulty with tasks including bending down without support or getting in and out of bed (Statistics New Zealand, 2014). It can be seen that the root of this issue is the ability to weight-bear and as such it is anticipated that limiting the amount of weight borne will increase the stability of a patient.

2.5.5 Cost

Research consistently documents the costs of patient and carer injuries greatly decrease when zero-force programs and assistive devices are introduced to facilities (Collins, Nelson, & Sublet, 2006; Accident Compensation Corporation, 2003; Borner, 2008). In 1999, nurses injured by patient handling in New Zealand cost over NZD$30 million (Accident Compensation Corporation, 2003). Garg (1999) found an average payback period of 15 months for nursing homes investing in "Zero-Lift Program" training and assistive lifting devices. Currently the acceptable cost range for a mechanical lift is between $3000 and $6000 (Collins, Nelson, & Sublet, 2006).

2.5.6 Manoeuvrability

A key reason carers choose not to use assistive lifting devices is their limited manoeuvrability (Li, Wolf, & Evanoff, 2004). It is specified that large items, such as beds should have a maximum turning circle of 1800 millimetre radius. For a smaller item, such as an independent wheelchair, this is decreased to 1500 millimetres (Accident Compensation Corporation, 2003). The storage ability of items must also be considered (Enos, 2008). The minimum door width within medical facilities is recommended to be 900 millimetres (Accident Compensation Corporation, 2003). As such, items should be adequately narrow to allow for easy storage and retrieval. It has been decided the turning circle of a device can be considered a fair representation of its manoeuvrability.

2.5.7 Cognitive Requirement

It is noted cognition is required in any lift where the patient must remain focused. Abusive behaviour directed at carers from patients is decreased when using assistive devices (Borner, 2008). Assessment of patients is required to gauge their cooperation, comprehension, and any barrier to completing a lift (Accident Compensation Corporation, 2003). This indicates that often, while a patient could physically use a simpler transfer aid, their cognition limits this.

2.5.8 Summary

Gauging device performance can be completed by reviewing the key requirements of an assistive lifting device. An evaluation matrix has been completed, shown in Table 1. When an assistive device is measured against this, the total value is referred to as the Design Specification Rating (DSR).

Table 1 Evaluation Matrix for Assistive Lifting Devices

Success Criteria / Value	**Ease of Use**	**Carer Input**	**Safety**	**Stability**	**Cost**	**Manoeuvrability**	**Cognitive Requirement**
	Time taken for transfer (TT)	Carer force as percentage of body weight (CF)	Number of the following criteria met (MC)	Percentage of weight patient-supported (WB)	Estimated cost of device in NZD (DC)	Turning circle radius (TC)	Patient cognitive requirement (CR)
1	$10\,min < TT$	$80\% < CF$	• No carer forward back bending • Load max 16kg • Locking mechanism • No twisting of carer torso • No dragging of patient • Force spread over large portion of patient's body • Varied surface height • No trap points • Transportable • Operation errors easily reversed	$100\% \leq WB$	$\$5000 \leq DC$	$2.2m \leq TC$	$100\%\ CR$
2	$8\,min < TT \leq 10\,min$	$70\% < CF \leq 80\%$		$80\% < WB < 100\%$	$\$4000 < DC < \$5000$	$2.1m < TC < 2.2m$	
3	$6\,min < TT \leq 8\,min$	$60\% < CF \leq 70\%$		$70\% < WB \leq 80\%$	$\$3000 < DC \leq \$4000$	$2.0m < TC \leq 2.1m$	Able to stay alert and focused for entire lift
4	$5\,min < TT \leq 6\,min$	$50\% < CF \leq 60\%$		$60\% < WB \leq 70\%$	$\$2000 < DC \leq \$3000$	$1.8m < TC \leq 2.0m$	
5	$4\,min < TT \leq 5\,min$	$40\% < CF \leq 50\%$		$50\% < WB \leq 60\%$	$\$1000 < DC \leq \$2000$	$1.6m < TC \leq 1.8m$	Able to follow a series of simple instructions
6	$3\,min < TT \leq 4\,min$	$30\% < CF \leq 40\%$		$30\% < WB \leq 50\%$	$\$750 < DC \leq \$1000$	$1.4m < TC \leq 1.6m$	
7	$2\,min < TT \leq 3\,min$	$20\% < CF \leq 30\%$		$20\% < WB \leq 30\%$	$\$500 < DC \leq \$750$	$1.2m < TC \leq 1.4m$	Able to follow a simple instruction
8	$1\,min < TT \leq 2\,min$	$10\% < CF \leq 20\%$		$10\% < WB \leq 20\%$	$\$250 < DC \leq \$500$	$1.0m < TC \leq 1.2m$	
9	$30\,s < TT \leq 1\,min$	$5\% < CF \leq 10\%$		$0\% < WB \leq 10\%$	$\$100 < DC \leq \$250$	$0.5m < TC \leq 1.0\,m$	$0\%\ CR$
10	$TT \leq 30\,s$	$CF \leq 5\%$		$WB \leq 0\%$	$DC \leq \$100$	$TC \leq 0.5m$	

2.6 Existing Solutions

Patient lifting and mobility equipment is recommended by the Accident Compensation Corporation (ACC) to limit the risk of injury to carers of low mobility patients. Equipment recommended by ACC for sitting transfers similar to the movements of the patient lifter includes Mobile Hoists, Standing Hoists, Transfer Belts, Transfer Boards, and Manual Lifters (Accident Compensation Corporation, 2012).

2.6.1 Mobile Hoist

A common piece of lifting and mobility equipment is the mobile hoist. These are the most complex of the discussed lifting and mobility equipment. Mobile hoists used in conjunction with patient slings attached to an overhead spreader bar. When the battery-operated hoist is activated, patients are lifted. Carer input is required to transfer the patient into the sling, attach the sling to the spreader bar, and, as mobile hoists are on casters, manoeuvre the loaded hoist. The benefit of this hoist is its ability to be used on fully dependant patients. Slings are available in a variety of configurations to allow maximum diversity of the hoist. One of the overriding limitations of a mobile hoist is the time taken to complete a lift. Alamgir et al. (2009) found that the average lift time for bed to chair transfers took 273.6 seconds, including 183.3 seconds to locate the hoist. Another limitation is the manual patient handling required to manoeuvre the patient into the sling before the lift.

Table 2 details the DSR of a mobile hoist where the maximum transfer force was taken to be 151.3 Newtons (Dutta, Holliday, Gorski, Baharvandy, & Fernie, 2012). It should be noted there is the possibility of carer forward back bending and carer torso twisting when loading the patient into the sling. The turning circle radius is taken from the Invacare Birdie Mobile Hoist and is 1400 millimetres (Invacare Corporation, 2009). The DSR of a mobile hoist is found to be 50, with the specification scores for each specification highlighted.

Table 2 Performance of Mobile Hoist

Success Criteria / Value	**Ease of Use**	**Carer Input**	**Safety**	**Stability**	**Cost**	**Manoeuvrability**	**Cognitive Requirement**
	Time taken for transfer (TT)	Carer force as percentage of body weight (CF)	Number of the following criteria met (MC)	Percentage of weight patient-supported (WB)	Estimated cost of device in NZD (DC)	Turning circle radius (TC)	Patient cognitive requirement (CR)
1	$10\ min < TT$	$80\% < CF$	• No carer forward back bending • **Load max 16kg** • **Locking mechanism** • No twisting of carer torso • **No dragging of patient** • **Force spread over large portion of patient's body** • **Varied surface height** • **No trap points** • **Transportable** • **Operation errors easily reversed**	$100\% \leq WB$	$\$5000 \leq DC$	$2.2m \leq TC$	$100\%\ CR$
2	$8\ min < TT \leq 10\ min$	$70\% < CF \leq 80\%$		$80\% < WB < 100\%$	$\mathbf{\$4000 < DC < \$5000}$	$2.1m < TC < 2.2m$	
3	$6\ min < TT \leq 8\ min$	$60\% < CF \leq 70\%$		$70\% < WB \leq 80\%$	$\$3000 < DC \leq \$4000$	$2.0m < TC \leq 2.1m$	Able to stay alert and focused for entire lift
4	$5\ min < TT \leq 6\ min$	$50\% < CF \leq 60\%$		$60\% < WB \leq 70\%$	$\$2000 < DC \leq \$3000$	$1.8m < TC \leq 2.0m$	
5	$\mathbf{4\ min < TT \leq 5\ min}$	$40\% < CF \leq 50\%$		$50\% < WB \leq 60\%$	$\$1000 < DC \leq \$2000$	$1.6m < TC \leq 1.8m$	Able to follow a series of simple instructions
6	$3\ min < TT \leq 4\ min$	$30\% < CF \leq 40\%$		$30\% < WB \leq 50\%$	$\$750 < DC \leq \$1000$	$1.4m < TC \leq 1.6m$	
7	$2\ min < TT \leq 3\ min$	$20\% < CF \leq 30\%$		$20\% < WB \leq 30\%$	$\$500 < DC \leq \$750$	$\mathbf{1.2m < TC \leq 1.4m}$	Able to follow a simple instruction
8	$1\ min < TT \leq 2\ min$	$\mathbf{10\% < CF \leq 20\%}$		$10\% < WB \leq 20\%$	$\$250 < DC \leq \$500$	$1.0m < TC \leq 1.2m$	
9	$30\ s < TT \leq 1\ min$	$5\% < CF \leq 10\%$		$0\% < WB \leq 10\%$	$\$100 < DC \leq \$250$	$0.5m < TC \leq 1.0\ m$	$\mathbf{0\%\ CR}$
10	$TT \leq 30\ s$	$CF \leq 5\%$		$\mathbf{WB \leq 0\%}$	$DC \leq \$100$	$TC \leq 0.5m$	

2.6.2 Standing Hoist

The standing hoist, like mobile hoists, utilise a sling and spreader bar and are predominantly battery operated. However, the sling of the standing hoist is fastened around the patient's torso, with knees being braced to knee supports via strapping while the patient is raised to standing. As a standing hoist requires patient input, patient dignity far outweighs that of the mobile hoist and it is the only device that aids in rehabilitating and increasing the mobility of users. It also has superior patient support to that of a transfer board or belt. Limitations of the standing hoist include the necessity for patients to be weight-bearing and the time taken for each lift.

Table 3 details the DSR of a standing hoist. It should be noted that maximum push force was taken to be 177.8 Newtons for a 90 kilogram patient (Lachance, et al., 2016). The maximum percent of patient weight borne was taken to be 79 percent (Tang, et al., 2016). Incorrect operation of the standing hoists can result in patients sliding out of slings (Garg, 1999). The turning circle radius is taken from the Invacare Roze and is 1380 millimetres (Invacare Corporation, 2014). The DSR of a standing hoist is found to be 36.

Table 3 Performance of Standing Hoist

<table>
<tr><th rowspan="2">Success Criteria / Value</th><th>Ease of Use</th><th>Carer Input</th><th>Safety</th><th>Stability</th><th>Cost</th><th>Manoeuvrability</th><th>Cognitive Requirement</th></tr>
<tr><th>Time taken for transfer (TT)</th><th>Carer force as percentage of body weight (CF)</th><th>Number of the following criteria met (MC)</th><th>Percentage of weight patient-supported (WB)</th><th>Estimated cost of device in NZD (DC)</th><th>Turning circle radius (TC)</th><th>Patient cognitive requirement (CR)</th></tr>
<tr><td>1</td><td>$10\ min < TT$</td><td>$80\% < CF$</td><td rowspan="10">• No carer forward back bending
• Load max 16kg
• Locking mechanism
• No twisting of carer torso
• No dragging of patient
• Force spread over large portion of patient's body
• Varied surface height
• No trap points
• Transportable
• Operation errors easily reversed</td><td>$100\% \leq WB$</td><td>$\$5000 \leq DC$</td><td>$2.2m \leq TC$</td><td rowspan="2">$100\%\ CR$</td></tr>
<tr><td>2</td><td>$8\ min < TT \leq 10\ min$</td><td>$70\% < CF \leq 80\%$</td><td>$80\% < WB < 100\%$</td><td>$\mathbf{\$4000 < DC < \$5000}$</td><td>$2.1m < TC < 2.2m$</td></tr>
<tr><td>3</td><td>$6\ min < TT \leq 8\ min$</td><td>$60\% < CF \leq 70\%$</td><td>$\mathbf{70\% < WB \leq 80\%}$</td><td>$\$3000 < DC \leq \$4000$</td><td>$2.0m < TC \leq 2.1m$</td><td rowspan="2">Able to stay alert and focused for entire lift</td></tr>
<tr><td>4</td><td>$5\ min < TT \leq 6\ min$</td><td>$50\% < CF \leq 60\%$</td><td>$60\% < WB \leq 70\%$</td><td>$\$2000 < DC \leq \$3000$</td><td>$1.8m < TC \leq 2.0m$</td></tr>
<tr><td>5</td><td>$\mathbf{4\ min < TT \leq 5\ min}$</td><td>$40\% < CF \leq 50\%$</td><td>$50\% < WB \leq 60\%$</td><td>$\$1000 < DC \leq \$2000$</td><td>$1.6m < TC \leq 1.8m$</td><td rowspan="2">Able to follow a series of simple instructions</td></tr>
<tr><td>6</td><td>$3\ min < TT \leq 4\ min$</td><td>$30\% < CF \leq 40\%$</td><td>$30\% < WB \leq 50\%$</td><td>$\$750 < DC \leq \$1000$</td><td>$1.4m < TC \leq 1.6m$</td></tr>
<tr><td>7</td><td>$2\ min < TT \leq 3\ min$</td><td>$\mathbf{20\% < CF \leq 30\%}$</td><td>$20\% < WB \leq 30\%$</td><td>$\$500 < DC \leq \$750$</td><td>$\mathbf{1.2m < TC \leq 1.4m}$</td><td rowspan="2">Able to follow a simple instruction</td></tr>
<tr><td>8</td><td>$1\ min < TT \leq 2\ min$</td><td>$10\% < CF \leq 20\%$</td><td>$10\% < WB \leq 20\%$</td><td>$\$250 < DC \leq \$500$</td><td>$1.0m < TC \leq 1.2m$</td></tr>
<tr><td>9</td><td>$30\ s < TT \leq 1\ min$</td><td>$5\% < CF \leq 10\%$</td><td>$0\% < WB \leq 10\%$</td><td>$\$100 < DC \leq \$250$</td><td>$0.5m < TC \leq 1.0\ m$</td><td rowspan="2">$0\%\ CR$</td></tr>
<tr><td>10</td><td>$TT \leq 30\ s$</td><td>$CF \leq 5\%$</td><td>$WB \leq 0\%$</td><td>$DC \leq \$100$</td><td>$TC \leq 0.5m$</td></tr>
</table>

2.6.3 Transfer Belt

Relevant transfer belts are defined as adjustable, padded belts that can be quickly attached and tightened around a patient's waist. Once fastened, carers can manoeuvre the patient using fabric handholds on the belt. These handholds are predominantly to aid the carer in stabilising the patient during the lift. This transfer aid is limited to weight-bearing, almost independent, cognitive patients, and is designed to assist rather than lift patients.

Table 4 details the DSR of a transfer belt. The maximum carer force is expected to be approximately 77 percent, as detailed further in Section 3.1. Tang et al. (2018) have found that carer forces can be as low as ten percent of the total patient weight for a correctly used walking belt with two carers. However, there are limitations to applying this finding to a standard sit-to-stand lift, including the use of able-bodied patients limited to 70 kilograms, the presence of carer coaching to ensure correct procedure was followed, and utilisation of two carers. It was decided the maximum force a carer would be expected to apply would be in the case of a patient falling, or being unable to hold their own weight. As such, the carer input is set at 77 percent of the patient's body weight. The DSR of a transfer belt is found to be 38.

Table 4 Performance of Transfer Belt

<table>
<tr><th rowspan="2">Success Criteria
Value</th><th>Ease of Use</th><th>Carer Input</th><th>Safety</th><th>Stability</th><th>Cost</th><th>Manoeuvrability</th><th>Cognitive Requirement</th></tr>
<tr><td>Time taken for transfer (TT)</td><td>Carer force as percentage of body weight (CF)</td><td>Number of the following criteria met (MC)</td><td>Percentage of weight patient-supported (WB)</td><td>Estimated cost of device in NZD (DC)</td><td>Turning circle radius (TC)</td><td>Patient cognitive requirement (CR)</td></tr>
<tr><td>1</td><td>$10\ min < TT$</td><td>$80\% < CF$</td><td rowspan="10">• No carer forward back bending
• Load max 16kg
• Locking mechanism
• No twisting of carer torso
• No dragging of patient
• Force spread over large portion of patient's body
• Varied surface height
• No trap points
• Transportable
• Operation errors easily reversed</td><td>$100\% \leq WB$</td><td>$\$5000 \leq DC$</td><td>$2.2m \leq TC$</td><td rowspan="2">$100\%\ CR$</td></tr>
<tr><td>2</td><td>$8\ min < TT \leq 10\ min$</td><td>$\mathbf{70\% < CF \leq 80\%}$</td><td>$\mathbf{80\% < WB < 100\%}$</td><td>$\$4000 < DC < \$5000$</td><td>$2.1m < TC < 2.2m$</td></tr>
<tr><td>3</td><td>$6\ min < TT \leq 8\ min$</td><td>$60\% < CF \leq 70\%$</td><td>$70\% < WB \leq 80\%$</td><td>$\$3000 < DC \leq \$4000$</td><td>$2.0m < TC \leq 2.1m$</td><td rowspan="2">Able to stay alert and focused for entire lift</td></tr>
<tr><td>4</td><td>$5\ min < TT \leq 6\ min$</td><td>$50\% < CF \leq 60\%$</td><td>$60\% < WB \leq 70\%$</td><td>$\$2000 < DC \leq \$3000$</td><td>$1.8m < TC \leq 2.0m$</td></tr>
<tr><td>5</td><td>$4\ min < TT \leq 5\ min$</td><td>$40\% < CF \leq 50\%$</td><td>$50\% < WB \leq 60\%$</td><td>$\$1000 < DC \leq \$2000$</td><td>$1.6m < TC \leq 1.8m$</td><td rowspan="2">Able to follow a series of simple instructions</td></tr>
<tr><td>6</td><td>$3\ min < TT \leq 4\ min$</td><td>$30\% < CF \leq 40\%$</td><td>$30\% < WB \leq 50\%$</td><td>$\$750 < DC \leq \$1000$</td><td>$1.4m < TC \leq 1.6m$</td></tr>
<tr><td>7</td><td>$2\ min < TT \leq 3\ min$</td><td>$20\% < CF \leq 30\%$</td><td>$20\% < WB \leq 30\%$</td><td>$\$500 < DC \leq \$750$</td><td>$1.2m < TC \leq 1.4m$</td><td rowspan="2">Able to follow a simple instruction</td></tr>
<tr><td>8</td><td>$\mathbf{1\ min < TT \leq 2\ min}$</td><td>$10\% < CF \leq 20\%$</td><td>$10\% < WB \leq 20\%$</td><td>$\$250 < DC \leq \$500$</td><td>$1.0m < TC \leq 1.2m$</td></tr>
<tr><td>9</td><td>$30\ s < TT \leq 1\ min$</td><td>$5\% < CF \leq 10\%$</td><td>$0\% < WB \leq 10\%$</td><td>$\$100 < DC \leq \$250$</td><td>$0.5m < TC \leq 1.0\ m$</td><td rowspan="2">$0\%\ CR$</td></tr>
<tr><td>10</td><td>$TT \leq 30\ s$</td><td>$CF \leq 5\%$</td><td>$WB \leq 0\%$</td><td>$\mathbf{DC \leq \$100}$</td><td>$\mathbf{TC \leq 0.5m}$</td></tr>
</table>

2.6.4 Transfer Board

In this research, the relevant transfer boards are defined as "sitting-to-sitting" boards. These are placed as a bridge between transfer surfaces and the patient slides over the board from one seat to the next. Users require cognition, stability, and strength. Table 5 details the DSR of a transfer board. Barbareschi, Cheng, and Holloway (2018) assessed the patient hand forces during transfer were 30 percent of body weight. It should be noted this only measured hand forces in patients with some leg weight-bearing ability. The total weight the patient would be required to support is expected to be approximately 77 percent, this is detailed further in Section 3.1. The DSR of a transfer board is 45.

Table 5 Performance of Transfer Board

<table>
<tr><th rowspan="2">Success Criteria
Value</th><th>Ease of Use</th><th>Carer Input</th><th>Safety</th><th>Stability</th><th>Cost</th><th>Manoeuvrability</th><th>Cognitive Requirement</th></tr>
<tr><td>Time taken for transfer (TT)</td><td>Carer force as percentage of body weight (CF)</td><td>Number of the following criteria met (MC)</td><td>Percentage of weight patient-supported (WB)</td><td>Estimated cost of device in NZD (DC)</td><td>Turning circle radius (TC)</td><td>Patient cognitive requirement (CR)</td></tr>
<tr><td>1</td><td>$10\ min < TT$</td><td>$80\% < CF$</td><td rowspan="10">• No carer forward back bending
• Load max 16kg
• Locking mechanism
• No twisting of carer torso
• No dragging of patient
• Force spread over large portion of patient's body
• Varied surface height
• No trap points
• Transportable
• Operation errors easily reversed</td><td>$100\% \leq WB$</td><td>$\$5000 \leq DC$</td><td>$2.2m \leq TC$</td><td rowspan="2">$\mathbf{100\%\ CR}$</td></tr>
<tr><td>2</td><td>$8\ min < TT \leq 10\ min$</td><td>$70\% < CF \leq 80\%$</td><td>$80\% < WB < 100\%$</td><td>$\$4000 < DC < \$5000$</td><td>$2.1m < TC < 2.2m$</td></tr>
<tr><td>3</td><td>$6\ min < TT \leq 8\ min$</td><td>$60\% < CF \leq 70\%$</td><td>$\mathbf{70\% < WB \leq 80\%}$</td><td>$\$3000 < DC \leq \$4000$</td><td>$2.0m < TC \leq 2.1m$</td><td rowspan="2">Able to stay alert and focused for entire lift</td></tr>
<tr><td>4</td><td>$5\ min < TT \leq 6\ min$</td><td>$50\% < CF \leq 60\%$</td><td>$60\% < WB \leq 70\%$</td><td>$\$2000 < DC \leq \$3000$</td><td>$1.8m < TC \leq 2.0m$</td></tr>
<tr><td>5</td><td>$4\ min < TT \leq 5\ min$</td><td>$40\% < CF \leq 50\%$</td><td>$50\% < WB \leq 60\%$</td><td>$\$1000 < DC \leq \$2000$</td><td>$1.6m < TC \leq 1.8m$</td><td rowspan="2">Able to follow a series of simple instructions</td></tr>
<tr><td>6</td><td>$3\ min < TT \leq 4\ min$</td><td>$30\% < CF \leq 40\%$</td><td>$30\% < WB \leq 50\%$</td><td>$\$750 < DC \leq \$1000$</td><td>$1.4m < TC \leq 1.6m$</td></tr>
<tr><td>7</td><td>$2\ min < TT \leq 3\ min$</td><td>$20\% < CF \leq 30\%$</td><td>$20\% < WB \leq 30\%$</td><td>$\$500 < DC \leq \$750$</td><td>$1.2m < TC \leq 1.4m$</td><td rowspan="2">Able to follow a simple instruction</td></tr>
<tr><td>8</td><td>$\mathbf{1\ min < TT \leq 2\ min}$</td><td>$10\% < CF \leq 20\%$</td><td>$10\% < WB \leq 20\%$</td><td>$\$250 < DC \leq \$500$</td><td>$1.0m < TC \leq 1.2m$</td></tr>
<tr><td>9</td><td>$30\ s < TT \leq 1\ min$</td><td>$5\% < CF \leq 10\%$</td><td>$0\% < WB \leq 10\%$</td><td>$\mathbf{\$100 < DC \leq \$250}$</td><td>$\mathbf{0.5m < TC \leq 1.0\ m}$</td><td rowspan="2">$0\%\ CR$</td></tr>
<tr><td>10</td><td>$TT \leq 30\ s$</td><td>$\mathbf{CF \leq 5\%}$</td><td>$WB \leq 0\%$</td><td>$DC \leq \$100$</td><td>$TC \leq 0.5m$</td></tr>
</table>

2.6.5 Manual Lifter

Conversely to mobile and standing hoists, manual lifters such as the Rand Scot Easy Pivot Lift use mechanical advantage rather than battery operation. These devices lever a patient from a sitting position to a forward leaning position. These lifts often use chest pads or lifting straps to lift the patient forwards out of a seat. With a smaller footprint and simpler mechanism to that of a mobile or standing hoist, these devices can be used quickly and effectively for suitable patients. The limitations present in a manual lifter are the need for patient stability and potentially high carer input forces. Table 6 details the DSR of a manual lifter. The DSR of a manual lifter is found to be 38.

Table 6 Performance of Manual Lifter

Value \ Success Criteria	Ease of Use	Carer Input	Safety	Stability	Cost	Manoeuvrability	Cognitive Requirement
	Time taken for transfer (TT)	Carer force as percentage of body weight (CF)	Number of the following criteria met (MC)	Percentage of weight patient-supported (WB)	Estimated cost of device in NZD (DC)	Turning circle radius (TC)	Patient cognitive requirement (CR)
1	$10\ min < TT$	$80\% < CF$	• No carer forward back bending • Load max 16kg • Locking mechanism • No twisting of carer torso • No dragging of patient • **Force spread over large portion of patient's body** • **Varied surface height** • **No trap points** • **Transportable** • **Operation errors easily reversed**	$100\% \le WB$	$\$5000 \le DC$	$2.2m \le TC$	$100\%\ CR$
2	$8\ min < TT \le 10\ min$	$70\% < CF \le 80\%$		$80\% < WB < 100\%$	$\$4000 < DC < \$5000$	$2.1m < TC < 2.2m$	
3	$6\ min < TT \le 8\ min$	$60\% < CF \le 70\%$		$\mathbf{70\% < WB \le 80\%}$	$\mathbf{\$3000 < DC \le \$4000}$	$2.0m < TC \le 2.1m$	Able to stay alert and focused for entire lift
4	$5\ min < TT \le 6\ min$	$50\% < CF \le 60\%$		$60\% < WB \le 70\%$	$\$2000 < DC \le \$3000$	$1.8m < TC \le 2.0m$	
5	$4\ min < TT \le 5\ min$	$40\% < CF \le 50\%$		$50\% < WB \le 60\%$	$\$1000 < DC \le \$2000$	$1.6m < TC \le 1.8m$	**Able to follow a series of simple instructions**
6	$3\ min < TT \le 4\ min$	$\mathbf{30\% < CF \le 40\%}$		$30\% < WB \le 50\%$	$\$750 < DC \le \$1000$	$1.4m < TC \le 1.6m$	
7	$\mathbf{2\ min < TT \le 3\ min}$	$20\% < CF \le 30\%$		$20\% < WB \le 30\%$	$\$500 < DC \le \$750$	$1.2m < TC \le 1.4m$	Able to follow a simple instruction
8	$1\ min < TT \le 2\ min$	$10\% < CF \le 20\%$		$10\% < WB \le 20\%$	$\$250 < DC \le \$500$	$\mathbf{1.0m < TC \le 1.2m}$	
9	$30\ s < TT \le 1\ min$	$5\% < CF \le 10\%$		$0\% < WB \le 10\%$	$\$100 < DC \le \$250$	$0.5m < TC \le 1.0\ m$	$0\%\ CR$
10	$TT \le 30\ s$	$CF \le 5\%$		$WB \le 0\%$	$DC \le \$100$	$TC \le 0.5m$	

2.6.6 Summary

Key specifications of existing devices are summarised in Table 7.

Table 7 Review of DSR Scores for Existing Assistive Devices

	Mobile Hoists	Standing Hoists	Transfer Belts	Transfer Boards	Manual Lifters
Ease of Use	5	5	8	8	7
Carer Input	8	7	2	10	6
Safety	8	7	3	5	5
Stability	10	3	2	3	3
Cost	2	2	10	9	3
Manoeuvrability	7	7	10	9	8
Cognitive Requirement	10	5	3	1	6
Total	**50**	**36**	**38**	**45**	**38**

From this, it can be seen that mobile hoists and transfer boards are the highest performing assistive devices when assessed with the developed evaluation matrix. The radar plot in Figure 1 is useful in assessing any gaps within the market.

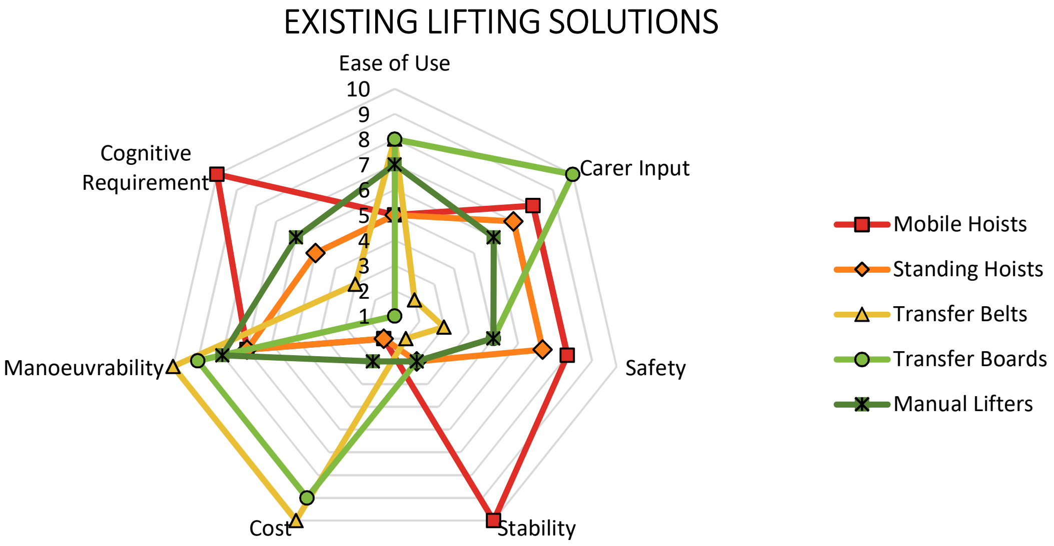


Figure 1 Summary of Existing Solutions Performance

It is clear that mobile hoists and transfer boards are effective transfer devices at either end of the market. Mobile hoists cater to patients with low mobility and cognition and focus on safety and support, sacrificing low cost and ease of use. Conversely, transfer boards provide a low cost product easy for both the patient and carer to use that require a patient to be relatively physically and mentally competent. Devices such as the manual lifters and standing hoists are required between these two extremes. The radar plots for both these devices are found to be less extreme with most of the DSR individual criteria rating between five and seven. Clearly, compared to mobile hoists and transfer boards, these devices are underperforming. From this, it is anticipated a device more user friendly and cheaper than a mobile hoist, but able to be used with less physically and mentally able patients than a transfer board, would be advantageous. It is expected that the device can be deemed successful if a DSR score of 45 or greater.

3 Theory

3.1 Patient Forces

No force, or minimal carer input force lifts, require the forces within the mechanism to be fully understood. To calculate the forces in the system it is first required to assess the forces applied to the device by the patient. Forces and angles are defined in the Nomenclature section. Approaching this from a statics perspective, from Newton's first law of motion, for the patient to remain in a defined position the forces must be at equilibrium. The force required to maintain the defined position is applied at the Alternate Lift Point (ALP). The ALP is defined as the point around which the patient rotates throughout the lift and is greatly dependant on the selected chest pad used. This is discussed further in Section 5.2. For a single pivot system with a passive chest pad, as shown in Figure 2, the ALP is defined as the centre point of the chest pad.

Figure 2 Single Pivot Lifter with Passive Chest Pad, ALP shown with ***o***

For a double pivot mechanism with an active chest pad, as shown in Figure 3, the ALP is defined as the pivot point. An active chest pad is defined as a chest pad the patient is rigidly attached to, causing the chest pad and the patient to move through the same angle throughout the lift. For this case, the ALP is defined as the pivot at which the Primary Pivot Arm and Secondary Pivot Arm meet.

Figure 3 Double Pivot Lifter with Active Chest Pad, ALP shown with ***o***

It is expected that the patient's knee position does not alter throughout the lifting process unless otherwise specified. The thigh segment has been modelled as a weightless link with vertical point forces applied at either end. The weight force is applied at the COM_T as shown in Figure 4 .

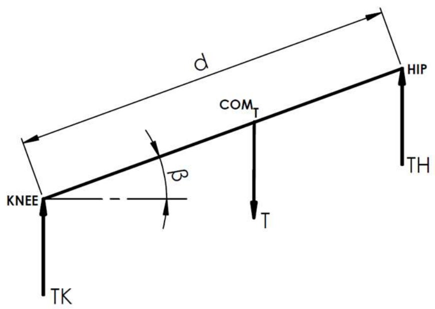


Figure 4 Point Forces on Patient's Thigh Segment

The magnitude of the point forces can be defined by Equations 1 and 2 for the knee and hip joints respectively. It should be noted that these are calculated through the vertical equilibrium equation and taking the moments around the knee.

$$\sum F_y = 0\text{: } TK + TH - T \qquad [1]$$

$$\Rightarrow TK = T - TH$$

$$\sum M_{knee} = 0\text{: } d.TH - \ COM_T.d.T \qquad [2]$$

$$TH = \frac{COM_T.d.T}{d}$$

$$\Rightarrow TH = \ COM_T.T$$

The lifted weight of the patient differs with patient body variations, ranging between 65 and 80 percent of the total body weight. A free body diagram of the total system, where the force applied at the ALP has been resolved into horizontal and vertical forces, CFx and CFy respectively, is shown in Figure 5.

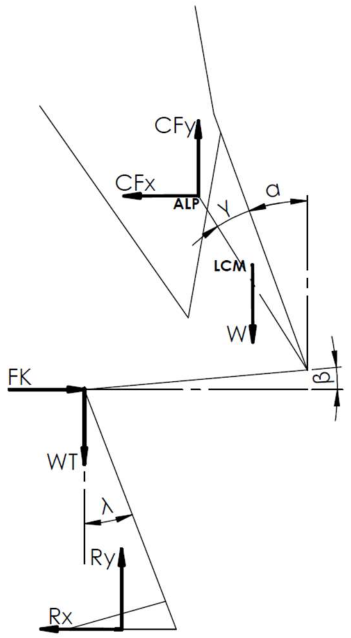


Figure 5 Total Patient Free Body Diagram

The equilibrium equations in the horizontal and vertical directions are shown below in Equations 3 and 4. Where the patient is held at a defined position, the total forces in both the horizontal and vertical directions should be equal to zero. It should be noted that WT is the reaction force to TK.

$$\sum F_x = 0\text{: } FK - Rx - CFx = 0 \qquad [3]$$

$$\Rightarrow Rx = FK - CFx$$

$$\sum F_y = 0\text{: } CFy - W - WT + Ry = 0 \qquad [4]$$

$$\Rightarrow Ry = W + WT - CFy$$

As the hip and knee joints can be considered pin joints, the patient's torso and thigh segments can be isolated and the equilibrium equations defined. It should be noted that, as pin joints, the knee and thigh can not support moments but can support vertical and horizontal forces. In reality, both joints can hold moments if applied by the patient although it is anticipated that any moments applied by the patient will decrease the carer input force by helping, rather than hindering, the lift. It is expected the highest carer input force will be required during a passive lift in which the patient applies no moments at the knee or hip. Figure 6 shows the isolated thigh segment and includes the internal thigh force TA and internal shank force FS.

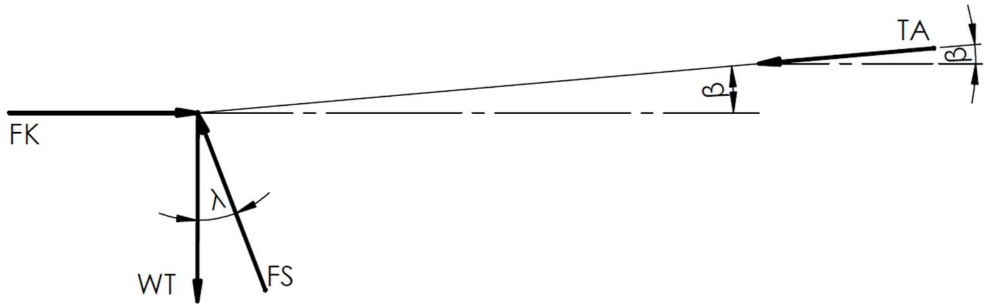


Figure 6 Thigh Free Body Diagram

The equilibrium equations for the thigh segment are shown in Equations 5 and 6 .

$$\sum F_x = 0:\ FK - TA\cos(\beta) - FS\sin(\lambda) = 0 \qquad [5]$$

$$\Rightarrow FK = TA\cos(\beta) + FS\sin(\lambda)$$

$$\sum F_y = 0:\ FS\cos(\lambda) - TA\sin(\beta) - WT = 0 \qquad [6]$$

$$\Rightarrow FS = \frac{TA\sin(\beta) + WT}{\cos(\lambda)}$$

Figure 7 shows the isolated torso segment.

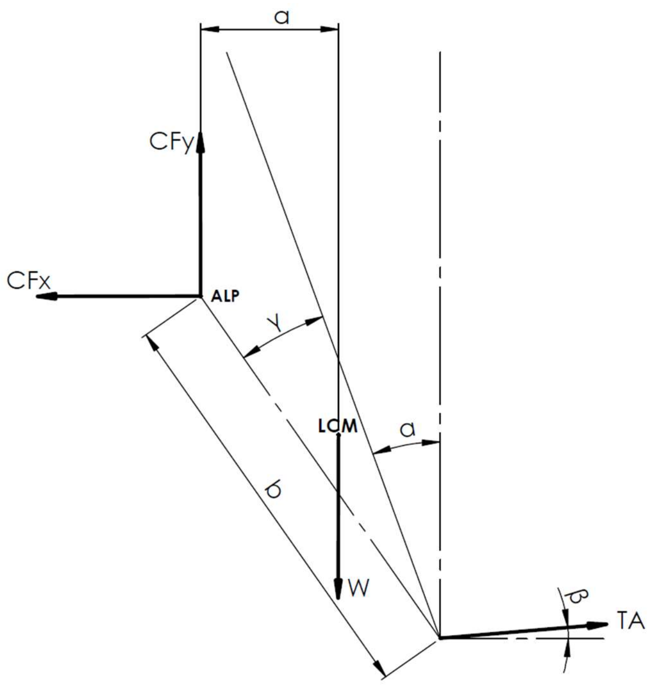


Figure 7 Torso Free Body Diagram

As the hip joint is modelled as a pin joint, a moment can not be supported. Therefore, the only force acting at the hip joint is TA, reversed as it is the reaction of TA shown in Figure 6. The equilibrium equations in the vertical and horizontal directions for the torso segment are shown in Equations 7 and 8.

$$\sum F_x = 0:\ CFx - TA\cos(\beta) = 0 \quad [7]$$
$$\Rightarrow CFx = TA\cos(\beta)$$

$$\sum F_y = 0:\ CFy - W + TA\sin(\beta) = 0 \quad [8]$$
$$\Rightarrow CFy = W - TA\sin(\beta)$$

The sum of the moments can be taken about the ALP. It should be noted only the force component of TA acting in the direction perpendicular to the ALP path, TR, will create a moment. This is shown in Figure 8.

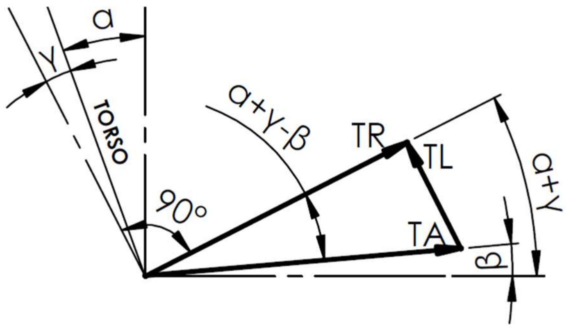


Figure 8 Component of Thigh Force Acting Perpendicular to ALP Line

TR is the component of TA that is perpendicular to the radius of the ALP. TA can be found using Equation 9.

$$TA = \frac{TR}{\cos(\alpha+\gamma-\beta)} \quad [9]$$

Taking moments about the ALP, the moment equilibrium equation is shown in Equation 10.

$$\sum M_{ALP} = 0:\ W.a - TR.b = 0 \quad [10]$$
$$\Rightarrow TR = \frac{Wa}{b}$$

Equation 10 can then be calculated for a defined patient's characteristics and substituted into the above equations. Sample calculations of this method are shown in Section 3.2. For clarity, the above equations are listed in the order required for solving.

$$TR = \frac{Wa}{b} \quad [10]$$

$$TA = \frac{TR}{\cos(\alpha+\gamma-\beta)} \quad [9]$$

$$CFy = W - TA\sin(\beta) \quad [8]$$

$$CFx = TA\cos(\beta) \quad [7]$$

$$FS = \frac{TA\sin(\beta)+WT}{\cos(\lambda)} \quad [6]$$

$$FK = TA\cos(\beta) + FS\sin(\lambda) \quad [5]$$

$$Ry = W + WT - CFy \quad [4]$$

$$Rx = FK - CFx \quad [3]$$

Given a certain patient's height, weight and centre of mass location, the vertical (CFy) and horizontal (CFx) forces required to keep the patient in a defined position can be calculated for a defined ALP position. CFx and CFy can then be combined to create a resultant force, FC, acting at angle δ from the vertical position through simple trigonometry as shown in Equations 11 and 12.

$$FC = \sqrt{CFx^2 + CFy^2} \quad [11]$$

$$\delta = \tan^{-1}\left(\frac{CFx}{CFy}\right) \quad [12]$$

From the conservation of energy, work is not completed when the motion of an object is perpendicular to the direction of the applied force. In application, the patient will be moved and supported by the ALP which will traverse some sort of arc. If this arc is built as a supporting rail along which the ALP runs, the equilibrating force, FEI, is then defined as the magnitude of the resultant horizontal and vertical forces, FC, acting along the trajectory of the ALP arc.

It should be noted that δ is negative, and therefore measured counter clockwise from the vertical axis, when CFx is acting in the negative direction. As the resultant force is acting at an angle of δ from the vertical and the angle of the direction perpendicular to the ALP trajectory is defined as ε, the angle between the two is defined in Equation 13.

$$\theta = |\varepsilon - \delta| \quad [13]$$

Figure 9 shows the component geometry of the system and the component of the resultant equilibrating input force that is necessary to hold the patient in the defined position.

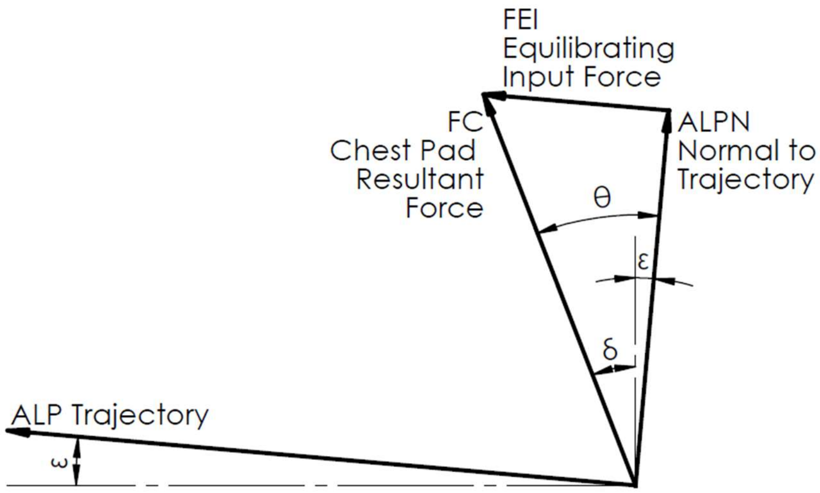


Figure 9 Definition of the Equilibrating Input Force

This force is defined in Equation 14.

$$FEI = FC\sin(\theta) \quad [14]$$

An example of the application of these calculations is shown in Section 3.2.

3.2 Patient Force Sample Calculations

In this section, the calculations discussed in Section 3.1 are completed for a patient with characteristics defined in Table 8.

Table 8 Sample Patient Characteristics

Characteristic	Value	Unit
Total Weight	70	kilograms
Percentage of total weight lifted	77.8	%
W Force	534	N
WT Force	152.4	N
A	0.1	m
B	0.49	m
α	13.7	deg
γ	11	deg
β	7.7	deg
ε	6.4	deg
λ	21	deg

Firstly, the moment in the torso will be addressed. A free body diagram of the torso segment can be seen previously in Figure 7, the moment equilibrium equation is shown in Equation 10. Equation 8 uses TR, defined previously in Figure 8.

$$TR = \frac{Wa}{b} \qquad [10]$$

$$\Rightarrow TR = \frac{534(0.1)}{0.49}$$

$$\Rightarrow \boldsymbol{TR = 109.0\ N}$$

As TR is the component of the TA force acting perpendicular to the ALP line, Equation 9 is used to define TA.

$$TA = \frac{TR}{\cos(\alpha+\gamma-\beta)} \qquad [9]$$

$$\Rightarrow TA = \frac{109.0}{\cos(13.7+11-7.7)}$$

$$\Rightarrow \boldsymbol{TA = 114.0\ N}$$

Equations 7 and 8 can then be used to solve the horizontal and vertical equilibrium equations for the torso segment.

$$CFx = TA\cos(\beta) \qquad [7]$$

$$\Rightarrow CFx = 114.0\cos(7.7)$$

$$\Rightarrow \boldsymbol{CFx = 113.0\ N}$$

$$CFy = W - TA\sin(\beta) \qquad [8]$$

$$\Rightarrow CFy = 534 - 114.0\sin(7.7)$$

$$\Rightarrow \boldsymbol{CFy = 518.7\ N}$$

A free body diagram of the thigh segment can be seen in Figure 6 previously, the equilibrium equations for the thigh segment are shown in Equations 5 and 6 . Equation 6 is completed first to provide a solution for FS.

$$\sum F_y = 0:\ FS\cos(\lambda) - TA\sin(\beta) - WT = 0 \qquad [6]$$

$$\Rightarrow FS = \frac{TA\sin(\beta) + WT}{\cos(\lambda)}$$

$$\Rightarrow FS = \frac{114.0\sin(7.7) + 152.4}{\cos(21)}$$
$$\Rightarrow \boldsymbol{FS = 179.6\ N}$$

$$FK = TA\cos(\beta) + FS\sin(\lambda) \tag{5}$$
$$\Rightarrow FK = 114.0\cos(7.7) + 179.6\sin(21)$$
$$\Rightarrow \boldsymbol{FK = 177.3\ N}$$

A free body diagram of the total system can be seen previously in Figure 5. The total patient equilibrium equations in the horizontal direction is shown in Equations 3 and 4.

$$Ry = W + WT - CFy \tag{4}$$
$$\Rightarrow Ry = 534 + 152.4 - 518.7$$
$$\Rightarrow \boldsymbol{Ry = 167.7\ N}$$

$$Rx = FK - CFx \tag{3}$$
$$\Rightarrow Rx = 177.3 - 113.0$$
$$\Rightarrow \boldsymbol{Rx = 64.3\ N}$$

Free body diagrams summarising these forces are shown in Figure 10.

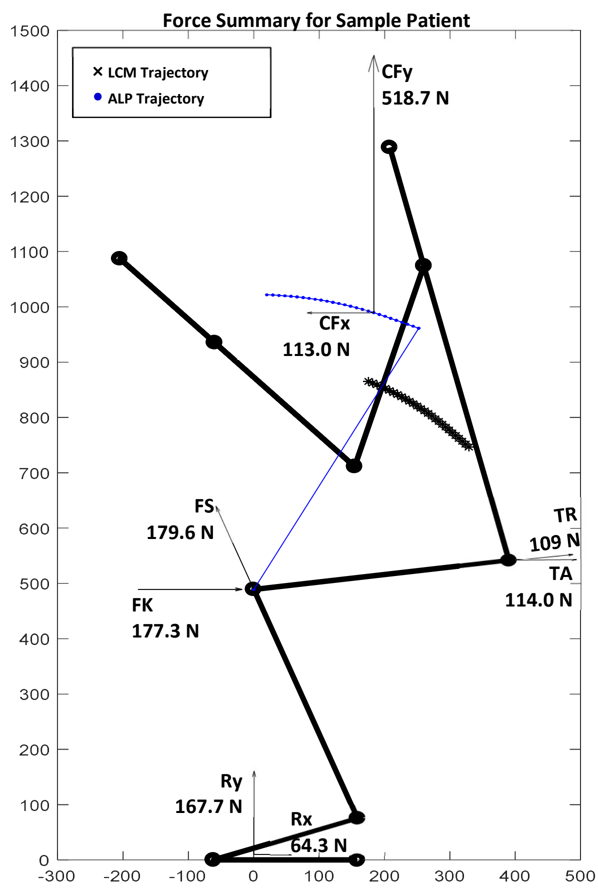


Figure 10 Force Summary for Sample Patient

CFx and CFy can then be combined to create a resultant force, FC, acting at angle δ from the vertical position through simple trigonometry as shown in Equations 11 and 12. It should be noted that as CFx is acting in the negative direction δ will be negative.

$$FC = \sqrt{CFx^2 + CFy^2} \quad [11]$$

$$FC = \sqrt{113^2 + 518.7^2}$$

$$FC = 530.9\ N$$

$$\delta = \tan^{-1}\left(\frac{CFx}{CFy}\right) \quad [12]$$

$$\delta = -\tan^{-1}\left(\frac{113}{518.7}\right)$$

$$\delta = -12.3\ ^{\circ}$$

From the conservation of energy, work is not completed when the motion of an object is perpendicular to the direction of the applied force. In the case where the force to hold the patient in place is applied directly at the ALP, the force is defined as the magnitude of the resultant horizontal and vertical forces, FC, acting along the trajectory of the ALP arc. As the resultant force is acting at an angle of δ from the vertical and the angle of the direction perpendicular to the ALP trajectory is defined as ε, the angle between the two is defined in Equation 13.

$$\theta = |\varepsilon - \delta| \quad [13]$$

$$\theta = |20.6 - -12.3|$$

$$\theta = 32.9^{\circ}$$

Figure 11 shows the component geometry of the system and the component of the resultant force that is necessary to hold the patient in the defined position for the sample case where the ALP trajectory is increasing in height at a 20.6 degree angle from the horizontal and the resultant chest pad angle is -12.3˚.

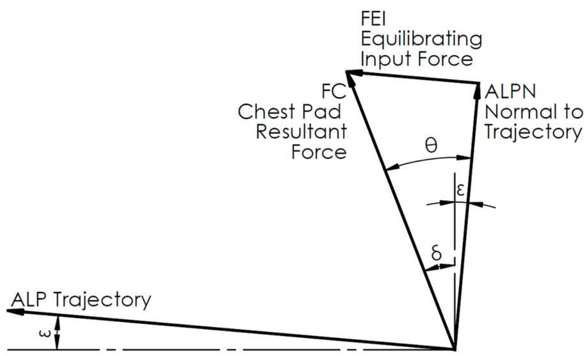


Figure 11 Definition of the Equilibrating Input Force

This force is defined in Equation 14.

$$FEI = FC\sin(\theta) \quad [14]$$

$$FEI = 530.9\sin(32.9)$$

$$\boldsymbol{FEI = 288.4\ N}$$

So, for the sample patient at the given position 288.4 Newtons of force, applied at the ALP along its trajectory, will keep the patient in the defined position.

3.3 No Input Force Case

Keeping forces reduced requires the understanding that the force required to hold a patient in a defined position is governed by the magnitude of the resultant chest pad force (FC) and the magnitude of this force along the trajectory of the ALP arc (FEI). From this, there are two ways to produce lower forces:

→ Lower the chest pad resultant force through reducing the total weight of patient
→ Reduce the magnitude of the resultant chest pad force not acting along the ALP trajectory

Whilst reducing the total weight of the system is not feasible, it is possible to reduce the magnitude of the equilibrating input force. This can be completed through decreasing the angle between the normal line to the ALP trajectory and the resultant chest pad force angle. From this, it can be expected that when the angle of the line normal to the ALP trajectory and the angle of the resultant chest pad force are equal, the equilibrating input force is equal to zero. This is shown in Equation 15.

$$FEI = FC\sin(\theta) = FC\sin(|\varepsilon - \delta|) \qquad [10]$$

$$\lim_{\varepsilon \to \delta} FEI = 0 \qquad [15]$$

The position and angle from the torso of the ALP relies heavily on the lifter mechanism and chest pad used for the lift. The ALP is defined as the point around which the torso rotates. For a simple mechanism, with a basic cylindrical chest pad, the patient's torso will rotate around the chest pad. For this case, the ALP point is defined as the centre point of the chest pad as shown in Figure 12.

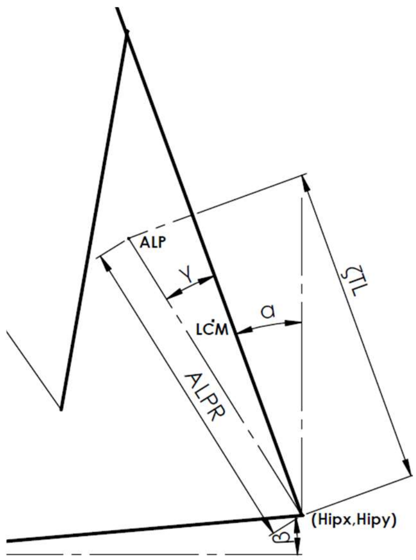


Figure 12 ALP Position Geometry

For cases where the ALP position is not impacted by the lifter mechanism the ALP position can be defined as in Equation 16, 17 and 18, chest pads are discussed in further in Section 5.2.

$$ALPR = \sqrt{\zeta {L_T}^2(1 + \tan^2\gamma)} \qquad [16]$$

$$ALPx = Hip_x - ALPR\cos(90 - \alpha - \gamma) \qquad [17]$$

$$ALPy = Hip_y + ALPR\sin(90 - \alpha - \gamma) \qquad [18]$$

For all cases, the angle from the horizontal to the ALP trajectory can be calculated using Equation 19. It should be noted that as this angle is perpendicular to the normal line, Equation 18 can also be used directly to calculate the angle of the ALP normal line from the vertical axis.

$$\varepsilon_n = \tan^{-1}\left(\frac{ALPy_n - ALPy_{n-1}}{ALPx_n - ALP\ \ _{n-1}}\right) \qquad [19]$$

A theoretical Zero Force ALP trajectory can be mapped for a specific patient. This can be completed iteratively by stepping through hip angles of the patient, iterating torso angles until δ and ε converge, and θ tends to zero. The Matlab code used is shown in Appendix A; a flow chart of the iteration process is included in Figure 13.

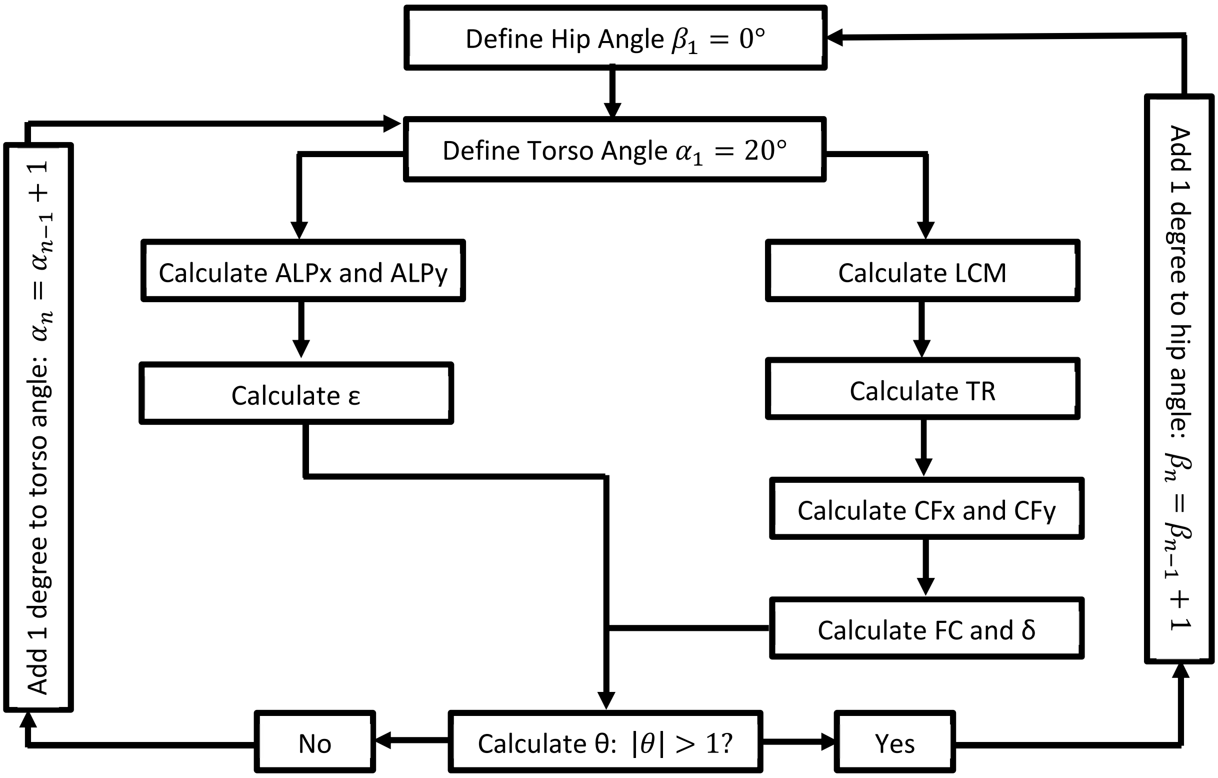


Figure 13 Flowchart of Iteration Process for Zero Force Simulation

To generate a physical zero force trajectory, a lifting mechanism would need to be developed that could follow the defined path. Options to create a zero force trajectory are discussed in Section 3.4.

3.4 Zero Force Case Approximation

It is expected the simplest method for approximating the defined zero force trajectory is with a mechanism defined by a single arc. The zero force trajectory for the sample patient was calculated, an arc was then fitted to this to calculate a radius and centre point. Firstly, the chord length of the trajectory and the angle of the chord from the vertical axis were calculated from the first and last ALP positions as shown in Equations 20 and 21.

$$CL_{ALP} = \sqrt{(ALPx_n - ALPx_1)^2 + (ALPy_n - ALPy_1)^2} \qquad [20]$$

$$\eta = \tan^{-1}\left(\frac{ALPx_n - ALPx_1}{ALPy_n - ALPy_1}\right) \qquad [21]$$

Once the chord height is found, the intersecting chord theorem can be used to calculate the radius and centre point of the arc. The chord height is found by calculating the difference between ALP chord midpoint and the ALP position which perpendicular bisector of the ALP chord. This point, ALP_{PB}, can be indicated as the only point where Equation 22 is true, Figure 14 shows the location of this point.

$$\tan(\eta) = \frac{ALPx - \ .5CLx_{ALP}}{ALPy \quad .5CLy_{ALP}} \tag{22}$$

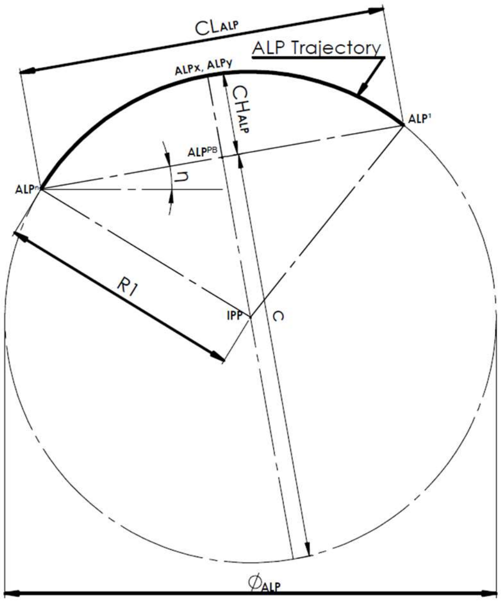


Figure 14 ALP Trajectory Chord Geometry

The chord height can then be calculated using the Intersecting Chords Theorem that when two chords intersect within a circle the product of their segments is equal, this is shown in Equation 23.

$$\frac{CL_{ALP}}{2} . \frac{CL_{ALP}}{2} = CH_{ALP} . c \tag{23}$$

$$\Rightarrow c = \frac{{CL_{ALP}}^2}{4CH_{ALP}}$$

A perpendicular bisector of a chord always passes through the centre of a circle, indicating that the diameter of the ALP trajectory, $\emptyset_{ALP}$, can be defined as in Equation 24.

$$\emptyset_{ALP} = CH_{ALP} + c \tag{24}$$

$$\Rightarrow \ \emptyset_{ALP} = CH_{ALP} + \frac{{CL_{ALP}}^2}{4CH_{ALP}}$$

Therefore, the radius and location of the primary pivot arm can be defined in Equations 25 to 27.

$$R1 = \frac{CH_{ALP}}{2} + \frac{{CL_{ALP}}^2}{8CH_{ALP}} \tag{25}$$

$$IPPx = ALPx + R1\sin(\eta) \tag{26}$$

$$IPPy = \ ALPy - R1\cos(\eta) \tag{27}$$

It should be noted that the radius and IPP will alter with varied patient characteristics. An approximation of the zero force path for the sample patient was completed and resulted in a radius of 0.98m and an IPP of (421, -37) from the origin.

3.5 Lifter Forces

The handle force can then be defined from the FEI and a defined handle geometry. For the theoretical single pivot as discussed above the mechanism forces can be calculated using a free body diagram. A free body diagram of the handle is shown in Figure 15.

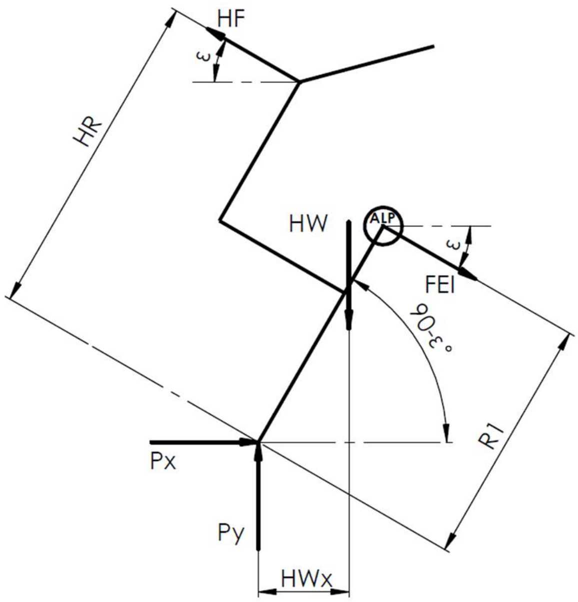


Figure 15 Free Body Diagram of Mechanism Handle

The horizontal and vertical equilibrium equations are shown in Equations 28 and 29.

$$\sum F_x = 0:\ Px + FEI\cos(\varepsilon) - HF\cos(\varepsilon) = 0$$
$$\Rightarrow Px = HF\cos(\varepsilon) - FEI\cos(\varepsilon) \qquad [28]$$

$$\sum F_y = 0:\ Py - HW - FEI\sin(\varepsilon) + HF\sin(\varepsilon) = 0$$
$$\Rightarrow Py = HW + FEI\cos(\varepsilon) - HF\sin(\varepsilon) \qquad [29]$$

It should be noted that the IPP joint is a pin joints and, as such, can not support moments. The moment equilibrium equation around the IPP is shown in Equation 30.

$$\sum M_{IPP} = 0:\ HWx.HW + R1.FEI - HR.HF = 0$$
$$\Rightarrow HF = \frac{HWx.HW + R1.FEI}{HR} \qquad [30]$$

These equations can then be used to calculate the necessary force the carer needs to apply to the handle and the resultant forces seen at the IPP. Calculating the carer force at each step of the iterative code discussed in Section 3.4 a force profile for the entire lift can be created.

Equations 26, 27 and 28 are used to calculate the handle force for the sample patient using the patient and lifter characteristics detailed in Table 9.

Table 9 Sample Patient and Lifter Characteristics

Characteristic	Value	Unit
Total Weight	70	kilograms
Percentage of total weight lifted	77.8	%
FEI Force	-13.9	N
HW	78.5	N
ε	22.8	Deg
R1	0.995	M
HWx	0.23	M
HR	1.15	M

The results of this are shown in Figure 16 with the Matlab code shown in Appendix F.

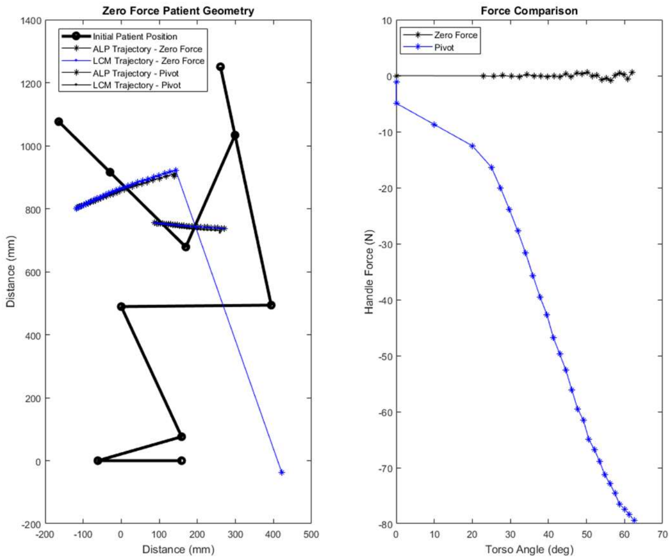


Figure 16 Single Pivot Approximation of Zero Force Lift, with Handle Weight Included

4 Patient Considerations

4.1 Overview

All discussion has focused on the assessment of forces and trajectories for a sample patient, the variation in patient shape and size and the implication of this should be considered. Obviously, patient height, weight, and centre of mass position should be considered and have a large impact on the forces and patient trajectories. It has been found that more subtle patient variations such as the thigh to torso length ratio also greatly affect the forces and trajectories. Access from Statistics New Zealand to the responses from the 2016 – 2017 New Zealand Health Survey was granted, allowing for assessment of the height, weight, waist circumference, and overall health of respondents. This has been used to provide clarity of the spread in patient characteristics that can be expected.

4.2 Weight

Basic physics indicates patient weight is an important component in assessing the handle forces present throughout a lift. It has been decided that the maximum allowable patient weight is 120 kilograms. Figure 17 details the outliers, 25th, 50th, and 75th percentiles for the weight of the total and aged 75 and over populations.

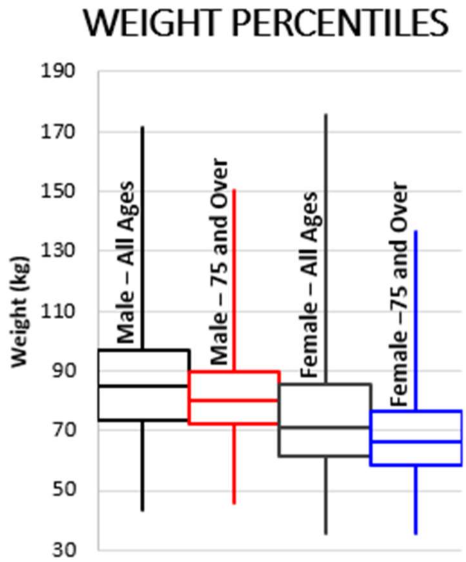


Figure 17 Box and Whisker Graph Indicating the Spread of Weight in the Total and Elderly Populations

From Health Data analysis, it was found that 94.6 percent of males and 96.7 percent of females within the total population were within the maximum weight criteria. It was also found that 98.9 percent of males and 99.5 percent of female aged 75 and over were within the maximum weight criteria. Figure 18 shows the normal distribution of weight of the New Zealand population.

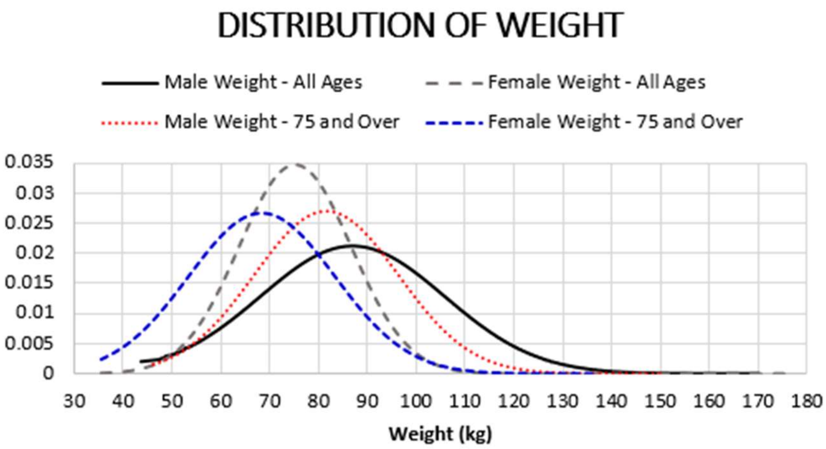


Figure 18 Distribution of Weight in New Zealand's Population, from New Zealand Health Data

It can be seen in the above graph that patient weight decreases with age, this occurs in both males and females. An interesting difference between males and females is that, while the spread of male weight in the 75 and over category decreases, the female weight spread does the opposite. It can be seen that the majority of male patients aged 75 and over weigh around 80 kilograms whereas the majority of females of the same age weigh between 55 and 75 kilograms. For calculations and MATLAB assessments, the 5th, 50th, and 95th percentiles were used. These are detailed in Table 10.

Table 10 Weight Values for Relevant Percentiles

Age Group	Gender	5th Percentile	50th Percentile	95th Percentile
All Ages	Male	60.9 kg	84.7 kg	121.4 kg
	Female	50.7 kg	71.0 kg	112.3 kg
Aged 75 and Over	Male	60.5 kg	80.0 kg	111.5 kg
	Female	46.5 kg	66.3 kg	95.7 kg

When a patient of constant height and varied weight is assessed, it is found that the peak handle force increases dramatically as shown in Figure 19.

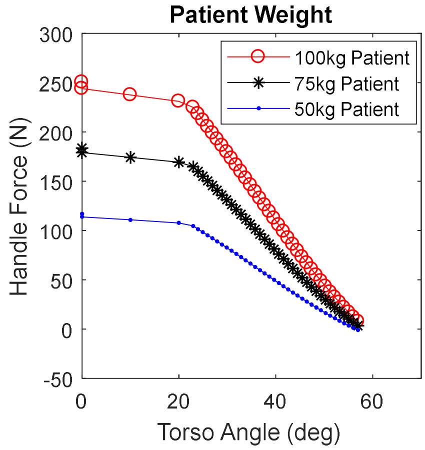


Figure 19 Handle Force for a Variety of Patient Weights

This is highly intuitive, for a heavier patient the force required to lift them will increase. The impact of weight distribution and its impact on patient location and trajectories are discussed further in Section 4.6.

4.3 Height

Evaluation of patient height was also completed using the 2016 – 2017 New Zealand Health Survey data. The main implication of patient height on handle forces is that the lifted centre of mass of a shorter patient is lower, resulting in a larger lift distance. Thus, a shorter person of larger weight can be found to have higher handle forces than a taller patient of the same weight for a non-adjustable mechanism. However, a shorter person will require an adjustable mechanism to be decreased in length to ensure contact with the kneepad. This adjustment will decrease handle force, and in some cases, cancel out the impact of a lower centre of mass. Figure 20 details findings of the height assessment.

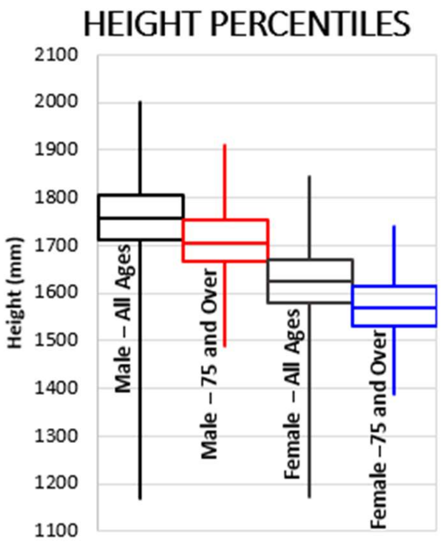


Figure 20 Box and Whisker Graph Indicating the Spread of Height in the Total and Elderly Populations

It was decided the maximum allowable height for patients would be 2000 millimetres. From Health Data analysis, it was found that 99.97 percent of males were within the maximum height criteria. It was also found that all persons aged 75 and over and all females were within the maximum height criteria. Figure 21 shows the normal distribution of height of the New Zealand population.

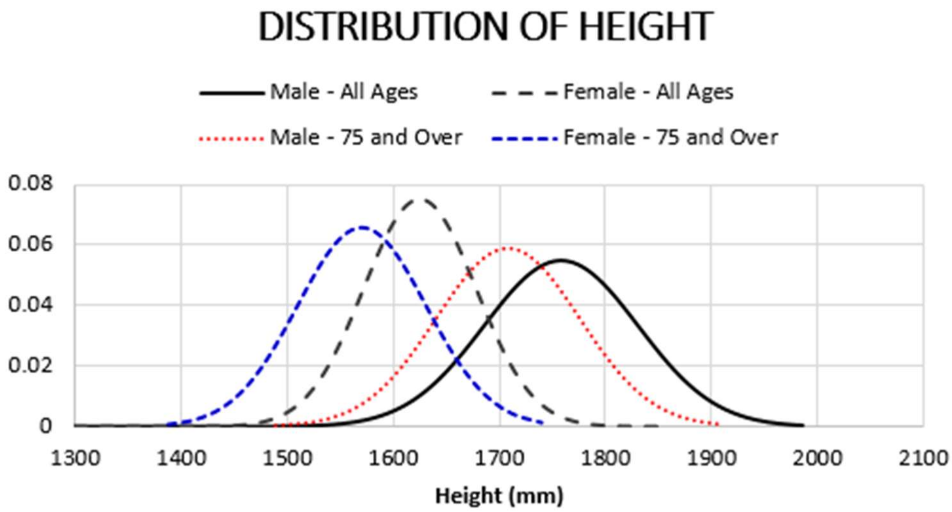


Figure 21 Distribution of Height in New Zealand's Population, from New Zealand Health Data

As expected, it is shown that patient height decreases with age. In both males and females, the decrease in the maxima of the bell curve is approximately 75 millimetres. It should be noted that the minimum height criteria varies between mechanisms. For calculations and MATLAB assessments, the 5th, 50th, and 95th percentiles were used. These are detailed in Table 11.

Table 11 Height Values for Relevant Percentiles

Age Group	Gender	5th Percentile	50th Percentile	95th Percentile
All Ages	Male	1642 mm	1759 mm	1881 mm
	Female	1511 mm	1626 mm	1732 mm
Aged 75 and Over	Male	1590 mm	1706 mm	1815 mm
	Female	1469 mm	1569 mm	1667 mm

The implications of patient height on handle force are intuitive. When a patient of constant weight and varied height is assessed, it is found that the peak handle force increases dramatically as shown in Figure 22.

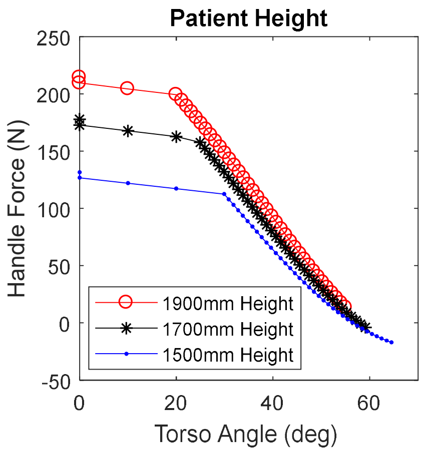


Figure 22 Handle Force for a Variety of Patient Heights

This is to be expected as, for a taller patient, the horizontal distance between LCM and ALP is increased. The longer shank length for a taller patient raises the knee, altering the patient's pivot point while the increased thigh length moves the patient's hip further away from the mechanism. The impact of patient height is shown in Figure 23 and is discussed further in Section 4.7.

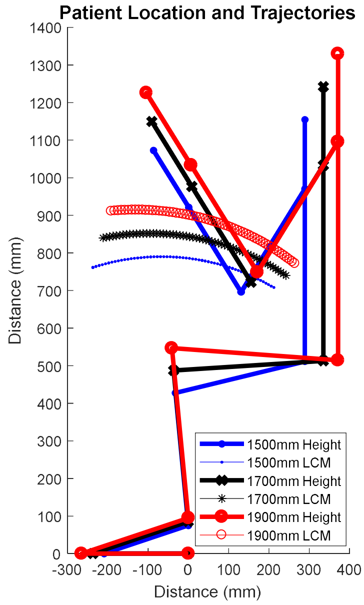


Figure 23 Patient Location and Trajectory Variation with Respect to Height

4.4 Waist Circumference

Circumference of the patient's waist was also assessed using the 2016 – 2017 New Zealand Health Survey data. This measurement was deemed an important factor in assessing the body type of patients. While waist circumference is highly impacted by weight, it is also valuable as a standalone measurement. Waist circumference can provide valuable information on the location of the lifted centre of mass. For a patient with a large waist circumference, the lifted centre of mass is moved further from the axis of the spine. This decreases the horizontal distance between the ALP and LCM, thus decreasing the value of TR. Figure 24 details the findings of the waist circumference assessment.

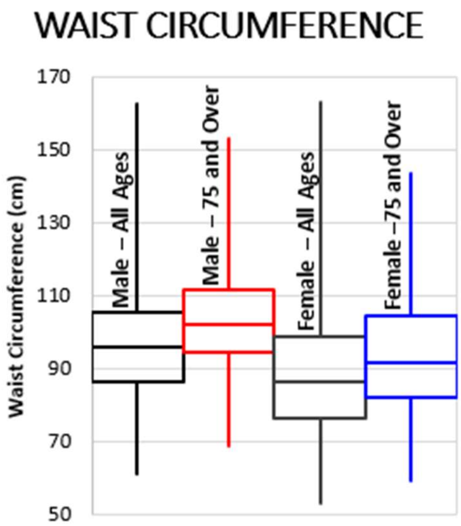


Figure 24 Spread of Waist Circumference in the Total and Elderly Populations

It was decided the maximum allowable waist circumference for patients would be 1200 millimetres. From Health Data analysis, it was found that 92.3 percent of males and 94.9 percent of females within the total population were within the maximum waist circumference criteria. It was also found that 91.5 percent of males and 95.6 percent of females aged 75 and over were within the maximum waist circumference criteria. Figure 25 shows the normal distribution of waist circumference of the New Zealand population.

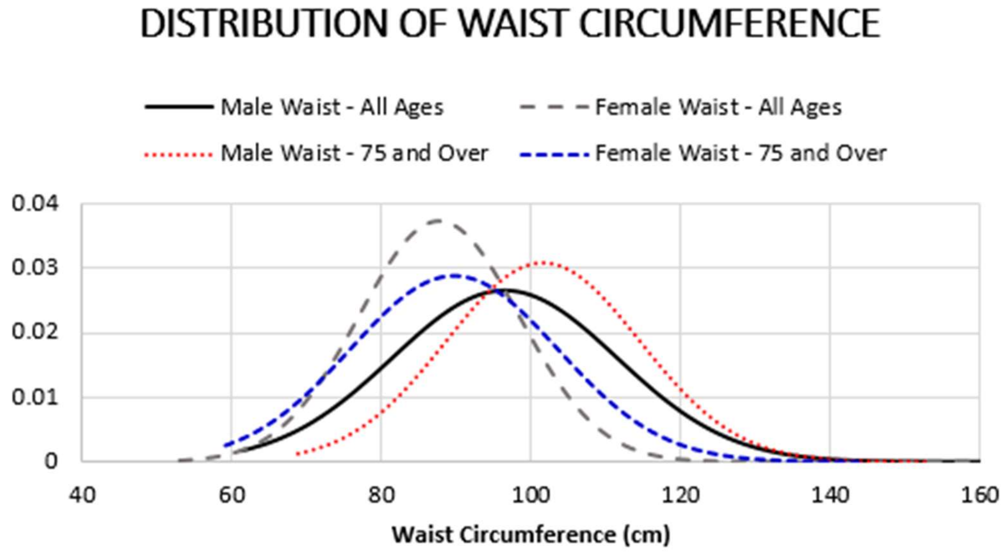


Figure 25 Distribution of Waist Circumference in New Zealand's Population, from New Zealand Health Data

It can be seen that the spread of waist circumference is larger in males than in females. For calculations and MATLAB assessment, the 5th, 50th, and 95th percentiles were used; these are detailed in Table 12.

Table 12 Waist Circumference Values for Relevant Percentiles

Age Group	Gender	5th Percentile	50th Percentile	95th Percentile
All Ages	Male	75.00 cm	95.95 cm	125.35 cm
	Female	66.65 cm	86.45 cm	120.25 cm
Aged 75 and Over	Male	81.55 cm	101.15 cm	123.45 cm
	Female	69.60 cm	90.00 cm	118.30 cm

4.5 Patient Mobility and Health Issues

Although the New Zealand Health Survey does not explicitly question the mobility of respondents, conclusions about the expected lifter market can be drawn. Individual questions have been analysed and combined to remove respondents outside the scope. Questions relevant to assessing patient mobility focussed on the self-reporting of respondents' overall health, how their health limited their activities, and daily exercise. This was used to develop an understanding of the market and the needs and characteristics of the aged 75 and over age group. Figure 26 shows how often in the last month respondents felt limited by their physical health when completing daily activities.

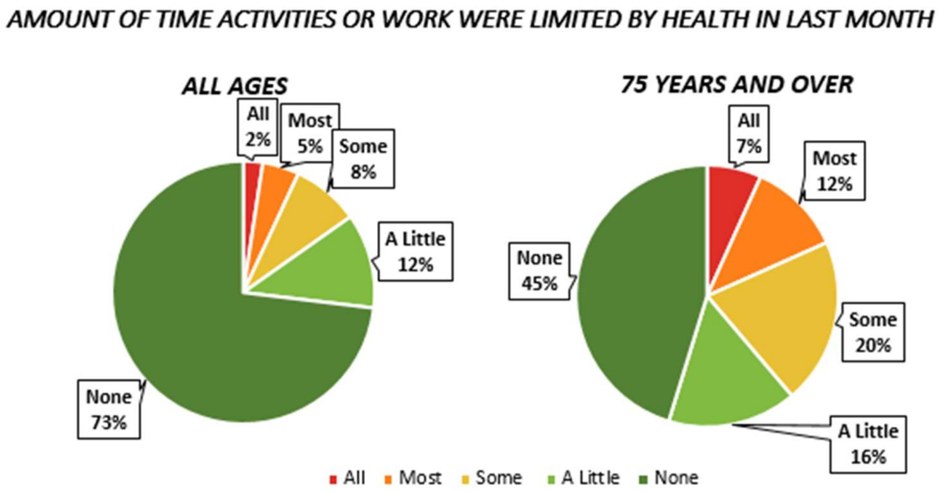


Figure 26 Comparison of All Ages and Aged 75 and Over Responses Regarding Health Limiting Activities

As is expected, there is a larger percentage of the respondents 75 and over reporting limitations in their lives due to their physical health. It is anticipated that respondents that report limitations all or most of the time would likely be limited in their mobility and benefit from a patient handling device. Figure 27 details responses regarding respondent's ability to climb stairs and the impact of their overall health on this.

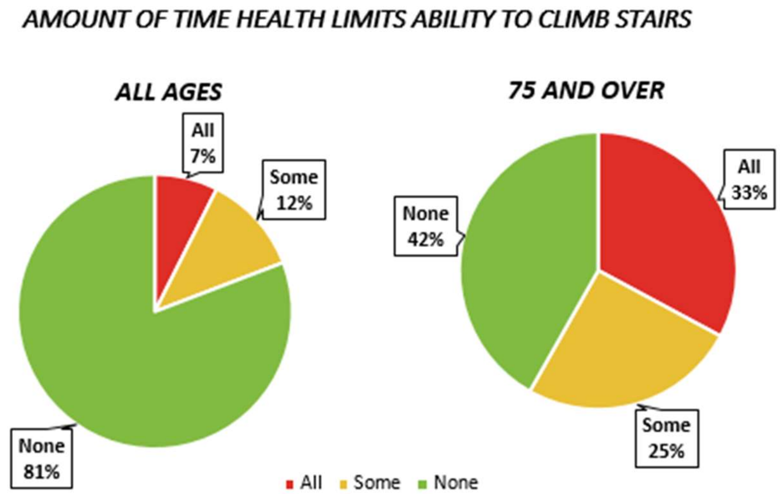


Figure 27 Comparison of All Ages and Aged 75 and Over Responses Regarding Health Limiting Stair Climbing

As is expected, there is a larger percentage of the respondents 75 and over reporting limitations to their stair climbing abilities due to their physical health. It is anticipated that respondents that report limitations all the time would likely by limited in their mobility or stability and benefit from a patient handling device. Figure 28 details the percentages of the population suffering from arthritis and to what extent.

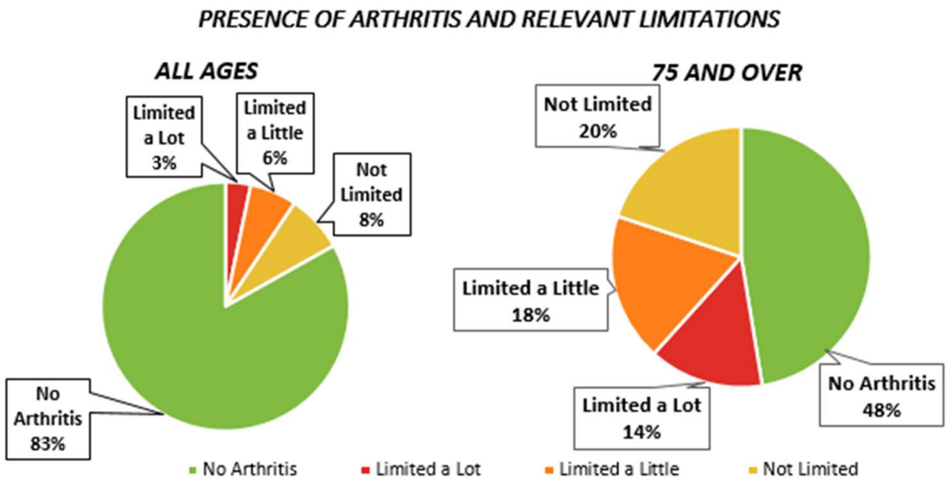


Figure 28 Comparison of Arthritis Present in Total Population and Aged 75 and Over Population

A larger percentage of the respondents 75 and over reported the presence of arthritis and limitations due to this. It is anticipated that respondents that report a lot of limitation would likely be limited in their mobility and may benefit from a patient handling device. Figure 29 shows the responses for the average amount of moderate physical activity completed per day.

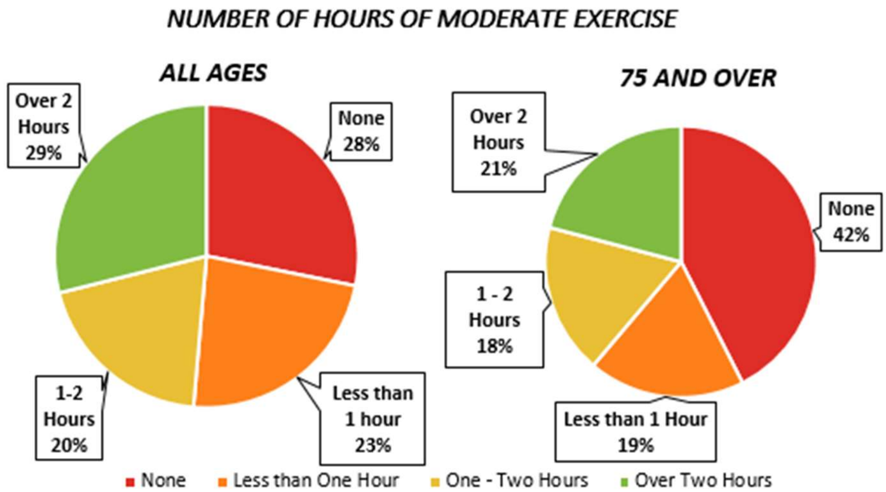


Figure 29 Comparison of All Ages and Aged 75 and Over Responses Regarding Moderate Physical Activity

A larger percentage of the respondents 75 and over reported no moderate physical activity. It is anticipated that respondents that reported no moderate physical activity would likely be limited in their mobility and may benefit from a patient handling device.

While mobility and stability are important criteria to assess when considering the market, more criteria can be applied to achieve further accurate representation. Using the New Zealand Health Survey, respondents were removed if they were deemed mobile through the above questions, exceeded the size and shape limits, living alone, pregnant, or their activities were limited by depression rather than physical health. Figure 30 shows the application of the criteria to assess the expected lifter market. It can be seen that the expected market for the lifter is 6.5 percent of the New Zealand population over the age of 15. It is anticipated that this is approximately 258,000 people.

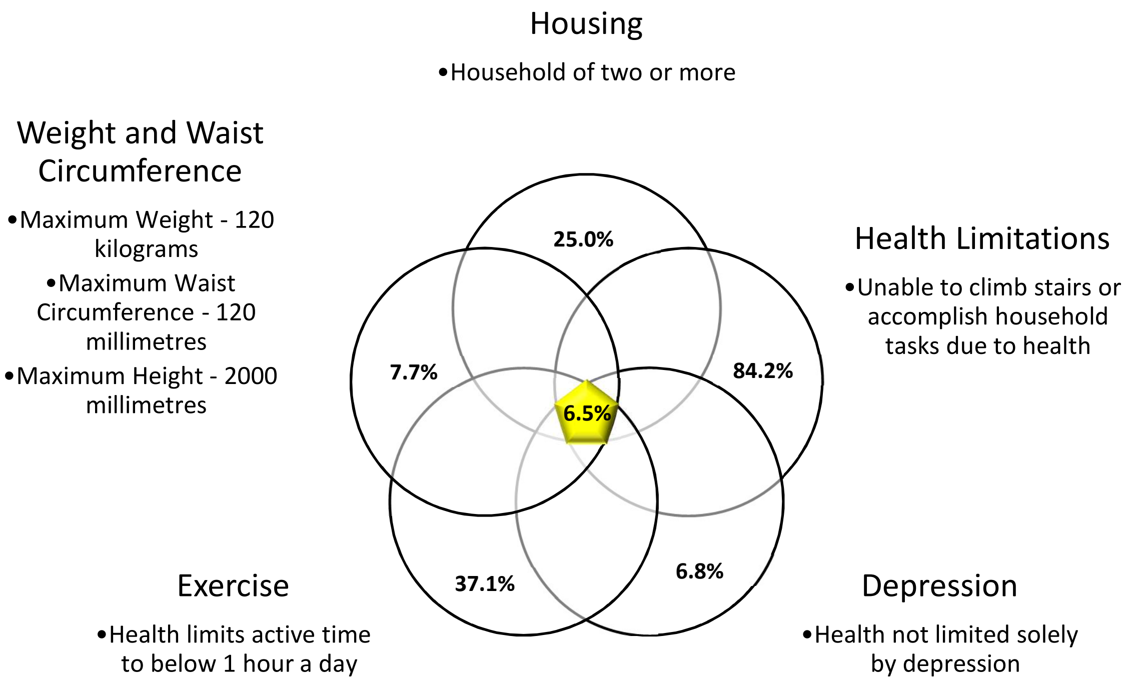


Figure 30 Venn Diagram Detailing the Criteria Used to Assess the Lifter Market

4.6 WEIGHT DISTRIBUTION

Understanding the implication of weight distribution is also important when assessing patient forces. Table 13 shows average height distributions and centre of mass locations from a variety of studies. Persons 5 and 6 show worst case scenarios developed to assess mechanism performance.

Table 13 Summary of Weight Distribution Variations

Percentage	Coefficient	PERSON	1	2	3	4	5	6	7	8
		Source	De Leva	Sample Patient	High LCM	Low LCM	Lower Body Dominant	Upper Body Dominant	De Leva	
									Elderly Female	Elderly Male
Mass (% of Total Weight)	C1	Head	8.3	8.1	11.0	6.0	8.3	8.3	7.39	6.94
	C2	Torso	46.8	49.7	53.0	30.0	46.8	46.8	46.22	51.52
	C3	Upper Arm	3.3	2.8	2.8	3.0	3.3	3.3	3.24	3.22
	C4	Forearm	1.9	1.6	1.6	1.0	1.9	1.9	1.12	1.76
	C5	Hand	0.7	0.6	0.6	0.5	0.7	0.7	0.50	0.65
	C6	Thigh	10.5	10.0	7.5	21.5	10.5	10.5	12.75	9.48
	C7	Shank	4.8	4.6	4.0	4.8	4.8	4.8	4.35	4.27
	C8	Foot	1.4	1.5	1.5	1.4	1.4	1.4	1.57	1.41
	C9	Total Lifted	77.1	62.2	81.5	66.5	77.1	77.1	77.14	79.25

Figure 31 outlines the implications of torso weight for the cases where the torso weight is 44.7 percent, 49.7 percent, and 54.7 percent of the patient weight.

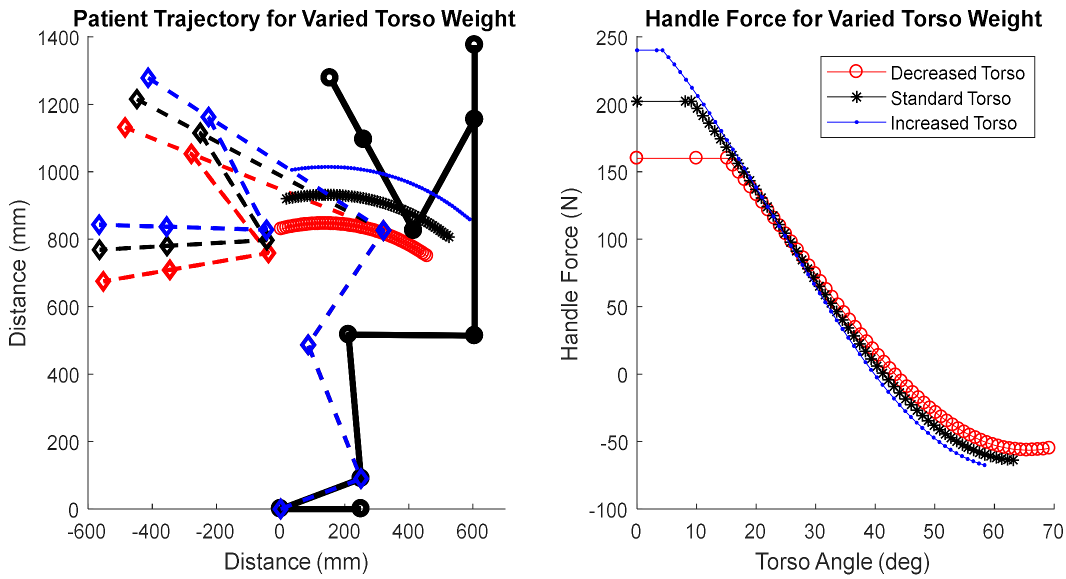


Figure 31 Implications of Torso Weight on Handle Forces and Trajectories

As expected, it can be seen that as the patient weight is increased, the handle force is increased. While this is partially due to the increase in total patient weight, it should be noted that adjusting the amount of torso weight respective to the total body weight also alters the position of the patient's LCM. It is noted that an increase in torso weight moves the LCM higher and further from the patient's knees. The same results can be seen when the thigh weight is altered. The case where the thigh weight is 5 percent, 10 percent, and 15 percent of the patient weight is shown in Figure 32.

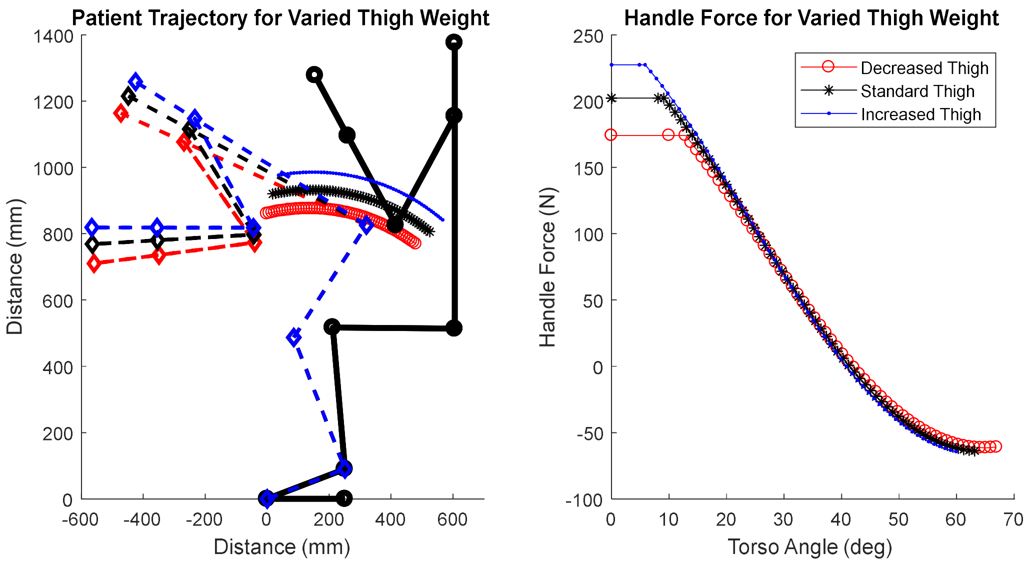
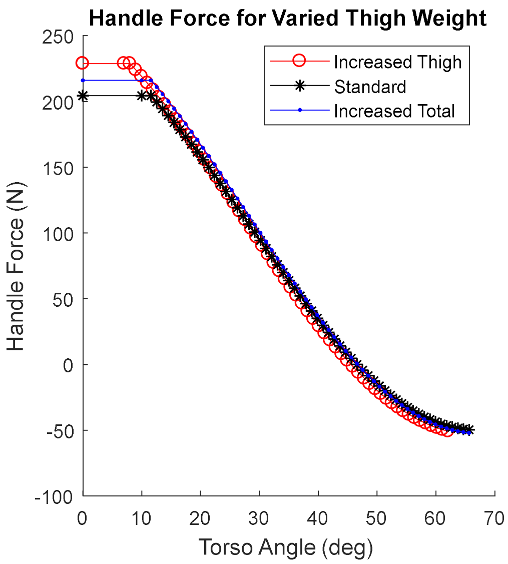


Figure 32 Implications of Torso Weight on Handle Forces and Trajectories

It can be seen that, similar to increasing the torso weight, the LCM is raised and moved closer to the patient's knees. It is expected that some of the variance in handle force is due to the change in total patient weight. However, when the increased torso values are compared to patients of respective increased weights, it can be see that increased torso and thigh weights more greatly influence the handle forces than evenly distributed weight. This is shown in Figure 33.



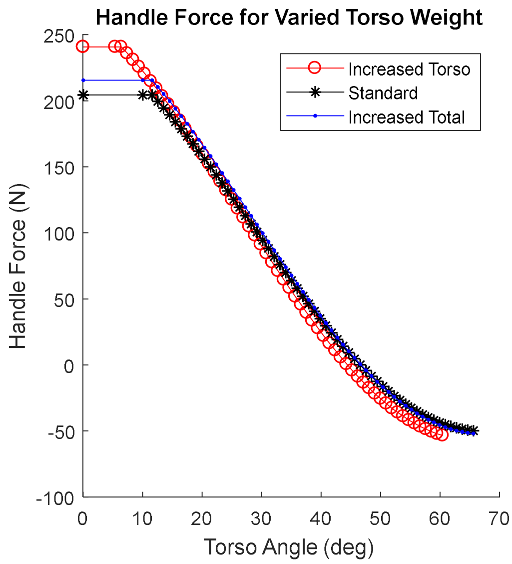


Figure 33 Comparison of Altered Body Segment Weights to Altered Total Weights

The impact of adjusted weight is especially noticeable in the torso. It is expected that this is due to the weight altering the position of the LCM. When a larger percentage of the total weight is located in the torso, the LCM is pushed upwards and further away from the patient's knee.

4.7 Height Distribution

Study of the height distribution is important when assessing patient forces. The implications of varied height distribution are complex and can be altered by a variety of factors. Table 14 shows average height distributions and centre of mass locations from a variety of studies. Persons 5 and 6 show worst case scenarios developed to assess mechanism performance during patient extremes.

Table 14 Summary of Height Distribution Variations

Percentage	Coefficient	PERSON	1	2	3	4	5	6	7	8
		Source	De Leva	Sample Patient	High LCM	Low LCM	Lower Body Dominant	Upper Body Dominant	De Leva Elderly Female	De Leva Elderly Male
Length (% of Total Height)	C10	Head	10.8	12.0	12.3	10.8	10.0	15.0	13.73	13.82
	C11	Torso	39.3	30.6	30.6	37.0	30.0	45.0	34.62	34.29
	C12	Upper Arm	17.2	21.1	21.1	17.2	17.0	18.0	16.85	16.85
	C13	Forearm	15.7	17.3	17.3	15.7	12.0	17.0	16.08	16.08
	C14	Hand	5.8	11.7	11.7	5.8	5.0	7.0	5.16	5.16
	C15	Thigh	23.2	22.0	22.0	23.2	27.0	20.0	21.71	24.47
	C16	Shank	24.7	24.8	24.8	24.7	30.0	17.0	25.47	25.15
	C17	Foot	4.3	4.3	4.3	4.3	4.3	3.0	15.44	15.44

Whilst the impact of overall height has been previously discussed in Section 4.3, the impact of height distribution has not been thoroughly assessed. Figure 34 details the impact on handle force when the shank length is altered; the cases considered are shank lengths of 20, 24.7, and 30 percent of total patient height. The following assessments are completed using the HTS3 mechanism.

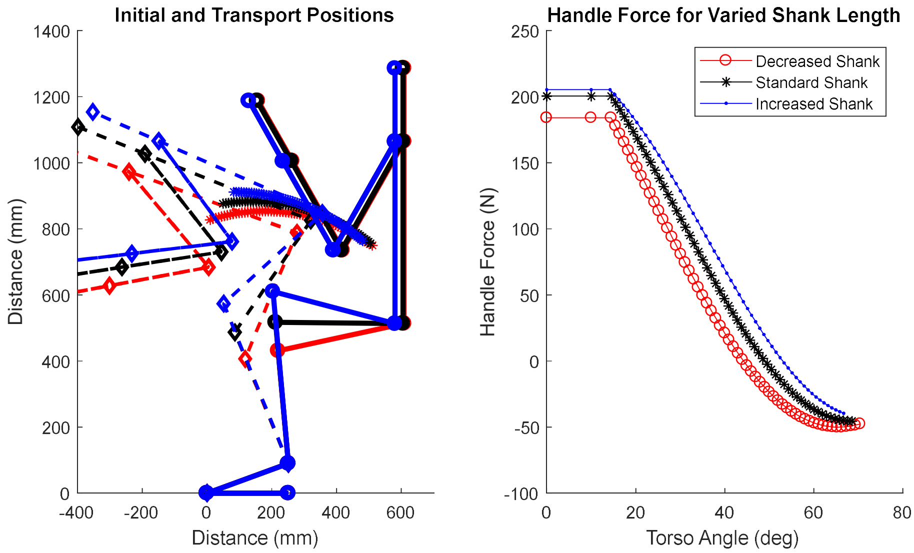


Figure 34 Implications of Shank Length on Handle Forces and Trajectories

From the above figure, it can be seen handle force increase is dependent on shank length. For the case where the shank length is equal to 30 percent of the patient height it is seen β is initially negative. Handle forces are reduced for all cases where the shank length is shorter than the height of the patient's seat. As such, instead of being dependant on the shank length, the handle force is impacted by the ratio between the seat height and the shank length. The implications of this ratio are shown in Figure 35 for the case where the shank length is 0.8, 1, and 1.2 times the height of the patient's seat.

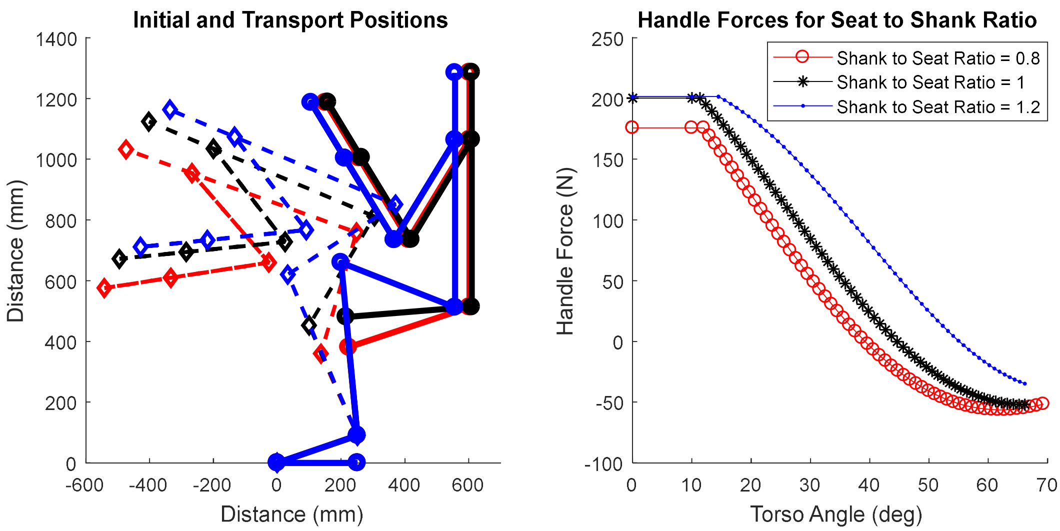


Figure 35 Implications of Shank to Seat Height Ratio on Handle Forces and Trajectories

As expected, forces are lower for the case where the shank to seat ratio is less than one. It is expected that this is due to the difference in thigh angle, altering the value of TA. The impact of thigh length was also considered; the cases where thigh length is 18 percent, 22 percent, and 26 percent of the total patient height are shown in Figure 36.

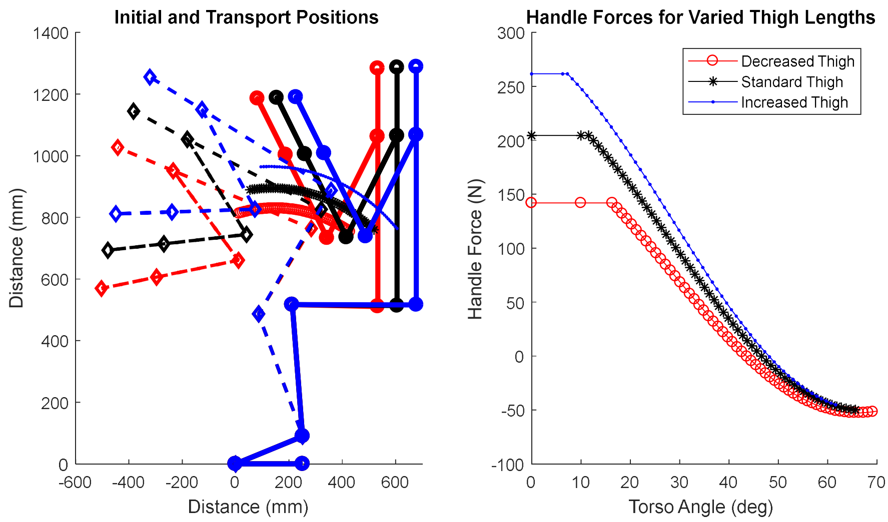


Figure 36 Implications of Thigh Length on Patient Position and Handle Forces

It is expected that the handle forces would be higher for increased thigh length. As discussed briefly in Section 3.1, this is due to the increased thigh length forcing the LCM further from the ALP. This is especially obvious at the initial lifting phase. It is anticipated that the higher initial forces for increased thigh length are also due to the increase in gradient of LCM trajectory during the initial stage of the lift. The gradient of the lift is increased due to the increase in the length of the pivot arm. Figure 37 outlines the implications of torso length for the cases where the torso length is 25.6 percent, 30.6 percent, and 35.6 percent of the patient height.

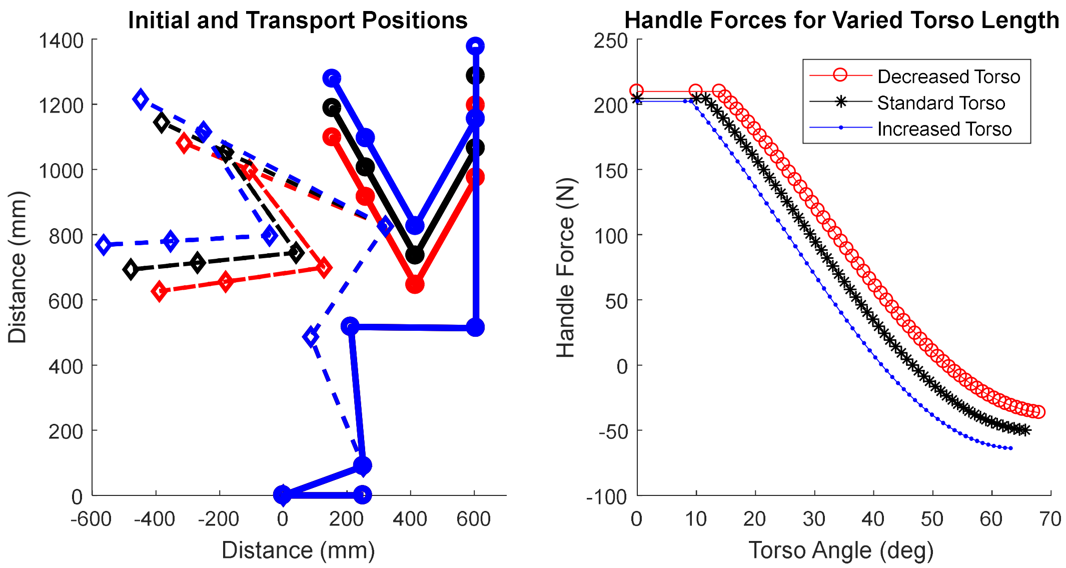


Figure 37 Implications of Torso Length on Patient Position and Handle Forces

From above, it can be seen that the torso height does not affect the initial handle forces. As the lift continues, however, the force is larger for a decreased torso length. This is due to the variance in torso angle increasing throughout the lift. At the transport position, it can be seen that a shorter torso results in an increased torso angle, increasing the horizontal distance between the ALP or VLP and the LCM. This results in an increase in handle force.

Although arm position and length will vary handle forces slightly, the impact of these on transport positions, trajectories, and handle force is considered negligible.

4.8 LIFTED CENTRE OF MASS

Evaluation of the patient's centre of mass is important when assessing the force applied to and by the patient when lifted. Table 15 details the centre of mass locations for body segments from a variety of studies.

Table 15 Summary of Centre of Mass for Body Segments

Percentage	Coefficient	PERSON	1	2	3	4	5	6	7	8
		Source	De Leva	Sample Patient	High LCM	Low LCM	Lower Body Dominant	Upper Body Dominant	De Leva	
									Elderly Female	Elderly Male
Centre of Mass Location*	C18	Head	55.0	44.0	65.0	35.0	55.0	55.0	51.8	54.0
	C19	Torso	63.0	41.0	75.0	15.8	63.0	63.0	51.9	52.7
	C20	Upper Arm	43.6	43.6	24.0	50.0	43.6	46.0	41.3	45.2
	C21	Forearm	43.0	40.0	60.0	40.0	43.0	50.0	53.3	58.0
	C22	Hand	46.8	50.6	50.0	35.0	46.8	43.6	22.8	22.4
	C23	Thigh	43.3	50.0	70.0	40.0	43.3	43.3	45.0	44.2
	C24	Shank	43.4	46.0	46.0	60.0	43.4	43.4	44.6	48.1
	C25	Foot	50.0	43.0	43.0	50.0	50.0	50.0	35.4	47.2

**Centre of mass percentage is located from the distal end of body segment*

The centre of mass locations for body segments can be combined with the height and weight distributions taken from Tables 13 and 14 to provide a position for the LCM. When assessing the forces and trajectories of the mechanisms, all simulations were completed using de Leva values for height and weight distributions and centre of mass positions. For patients where the LCM greatly differed from the de Leva values, an adjustment was made to the positioning of the LCM.

If patient's position is known, the LCM position can be assessed by measuring kneepad forces during a lift. Motion capture of a known mechanism can be used to assess the relevant patient angles, and the force applied to the footplate for a patient of known height and weight can be measured at two points along the lift. Equation 6 can be substituted into Equation 5.

$$FK = TA\cos(\beta) + FS\sin(\lambda) \qquad [5]$$

$$FS = \frac{TA\sin(\beta)+WT}{\cos(\lambda)} \qquad [6]$$

$$FK = TA\cos(\beta) + \frac{TA\sin(\beta)+WT}{\cos(\lambda)}\sin(\lambda) \qquad [31]$$

Equation 31 can then be rearranged in terms of TA as shown in Equation 32.

$$TA = \frac{FK\cos(\lambda)-WT\sin(\lambda)}{\cos(\beta)\cos(\lambda)+\sin(\beta)\sin(\lambda)} \qquad [32]$$

Equations 9 and 10 can then be substituted into Equation 32 as shown in Equation 33.

$$\frac{TR}{\cos(\alpha+\gamma-\beta)} = \frac{FK\cos(\lambda)-WT\sin(\lambda)}{\cos(\beta)\cos(\lambda)+\sin(\beta)\sin(\lambda)}$$

$$\frac{\frac{Wa}{b}}{\cos(\alpha+\gamma-\beta)} = \frac{FK\cos(\lambda)-WT\sin(\lambda)}{\cos(\beta)\cos(\lambda)+\sin(\beta)\sin(\lambda)}$$

$$a = \frac{\{FK\cos(\lambda)-WT\sin(\lambda)\}.\mathrm{b}.\cos(\alpha+\gamma-\beta)}{W\{\cos(\beta)\cos(\lambda)+\sin(\beta)\sin(\lambda)\}} \qquad [33]$$

From the calculation of the horizontal distance between the ALP or VLP and the LCM, a, at two points, the position of a patient's LCM could be calculated using the intersection of the two lines of possible locations of LCM. The equations for these lines can be calculated using the standard straight line equation y=mx+c where m and c are defined in Equations 34 and 35 respectively.

$$m = \tan(90 - \theta) \tag{34}$$

$$c = b - a \tag{35}$$

The location of a is given by the point where the Equation 36 is true.

$$m_1 x + (b - a_1) = m_2 x + (b - a_2) \tag{36}$$

5 Mechanism Considerations

5.1 Practical Parameters

Initial considerations must be completed before assessing pivot placement, the parameters of the lift need to be discussed. The pivot position should not collide with either the patient or lifting surface throughout the lift, and the mechanism should retain its functionality and lightweight characteristics. From these constraints, zone limitations were enforced. It was decided that pivot points behind the heel of the patient would not be considered (Zone A). Also, to maintain functionality of the device, the pivot location is limited to below the line between the position of the jugular notch at the initial position and the point at floor level 600 millimetres towards the carer from the patient's knee joint (Zone B). To ensure the mechanism does not collide with the patient, the pivot location is constrained to be below the line passing through the centre of mass of the thigh at a 20 degree angle from the horizontal (Zone C). To provide clearance and manoeuvrability, the pivot point must not be closer than 75 millimetres from the floor (Zone D). Finally, to ensure the pivot point does not collide with the carer the location is limited to 400 millimetres towards the carer from the patient's knee joint (Zone E). Any point not included in Zones A, B, C, D or E is deemed as a suitable location for a pivot point to be positioned as shown in Figure 38.

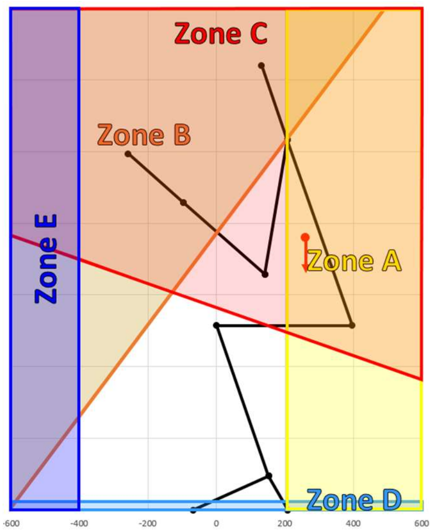


Figure 38 Suitable Pivot Point Zone

Section 3.4, discusses the approximation of a zero force lift by a single pivot mechanism. The pivot point discussed falls within Zone A and D which, in practice, would collide with the patient's chair and the floor, limiting practicality. Given this, it was decided to sacrifice the idea of a zero force lift and focus on delivering a low force lift utilising a simple single pivot mechanism.

When considering specific geometries for a mechanism it is also important that the trajectory of the patient throughout the lift is assessed. The angle between the thigh and torso should not decrease to a point where this causes patient discomfort. It was decided that the torso to thigh angle would be limited to an angle no smaller than 70 degrees. It is also important to ensure that the head does not drop below the heart at any point throughout the lift and the angle of the ankle does not exceed 20 degrees. The patient should be raised high enough that the transport position is stable. The mechanism can be considered stable when the LCM of the patient is located within the area encompassed by the mechanism wheels. The patient should also be easily placed on a variety of seating surfaces between 410 and 600 millimetres.

5.2 Chest Pad Implications

Noted previously in Section 3.1, two types of chest pads were considered, active and passive. A passive chest pad is defined as a chest pad that allows the torso of the patient to rotate independently of the rotation of the mechanism. A patient using a passive chest pad must have the strength and cognitive awareness to hold the handles and ensure there is no movement in their shoulder joint relative to their torso. The issues with a passive chest pad arise when a patient does not have the strength or cognition to aid the lifting process.

It was decided that an active chest pad would address the safety and functionality issues that arise with a passive chest pad. An active chest pad is defined as a chest pad that "captures" and holds the patient steady throughout the lift. It is anticipated that a fully active chest pad will allow for lifting of a patient without any patient input. It is important to note that an active chest pad can only be used on a mechanism where the rotation angle of the mechanism is equal to the required change in torso angle of a patient.

As each of the chest pad interacts differently with a patient, the ALP and contact points also change. For passive chest pads, the ALP can be defined as the centre point of the chest pad profile as this is the point around which the torso will rotate throughout the lift. For a case where there is more than one pivot, the term VLP is used to define to contact point between the patient and the chest pad. This is discussed in more detail in Section 8.3. The X and Y columns of Table 16 indicate how far the ζTL point is located from the ALP or VLP.

Table 16 Chest Pad Characteristics

Chest Pad	Active	ζ	X (mm)	Y (mm)	CPW (kg)
Standard	Passive	0.8	-100	-100	2
Large Blue	Passive	0.7	-125	-200	3
Bull Horns	Passive	0.8	-100	-75	1.5
Hug	Active	0.5	-50	-50	3
Pink	Passive	0.85	-25	-50	1

For clarity, Figure 39 shows the ALP and VLP points marked on an example chest pad for a single pivot mechanism. The handling of the chest pads within simulation is included in Appendix D.

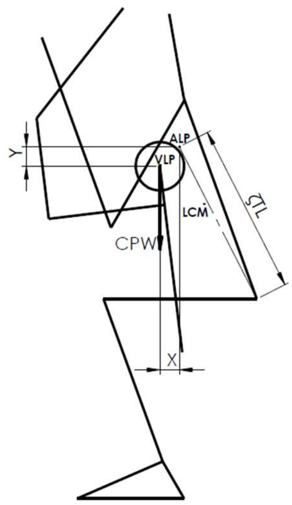


Figure 39 ALP Position Adjusted for Chest Pads

6 Testing Methods

6.1 Quasi-Static Handle Force Testing

Evaluation of handle forces was required to provide initial results to validate calculations. A spring balance was attached to the carer handle to measure carer forces. The setup for this testing is shown in Figure 40.

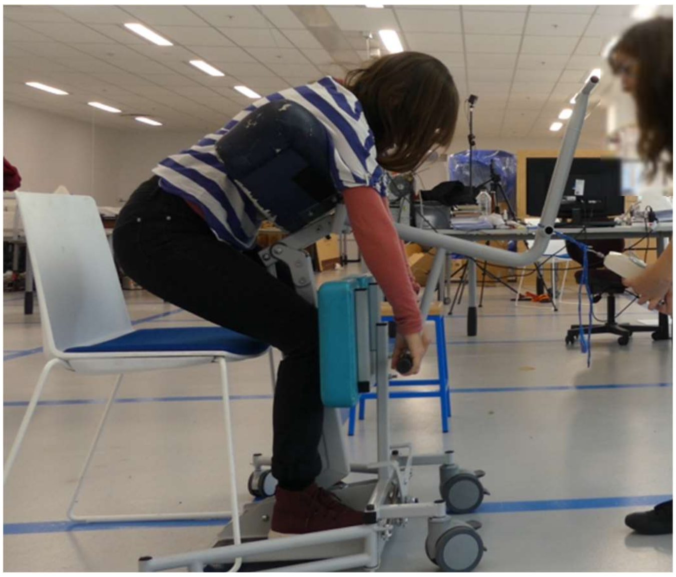


Figure 40 Quasi-Static Handle Force Testing

The method for quasi-static handle testing is detailed below:

1. Volunteer height and weight measured
2. Chair height measured
3. Lifter mechanism and configuration noted
4. Lift and test procedure discussed
5. Patient fully lifted and placed back on seat – control lift
6. Patient lifted to the point where the thigh loses contact with chair

7. Patient held steady at this point, ensuring no momentum present and torso angle noted
8. Force balance measured handle force, perpendicular to the ALP trajectory at this point
9. Patient lifted until torso angle was 45 degrees from vertical
10. Patient held steady at this point, ensuring no momentum present
11. Force balance measured handle force, perpendicular to the ALP trajectory at this point
12. Patient lifted until torso angle was 60 degrees from vertical
13. Patient held steady at this point, ensuring no momentum present
14. Force balance measured handle force, perpendicular to the ALP trajectory at this point
15. Patient lowered onto seat and released from any chest straps

The key limitation with the quasi-static testing method was the torso angle approximation. This technique was very inaccurate and resulted in a large level of uncertainty in test results. However, quasi-static testing did most accurately mirror the developed theory and was useful in initial validation of the theory and simulated models.

6.2 Dynamic Handle Force Testing

Validation of the total force profile and increased accuracy was achieved through dynamic handle force testing using a load cell. The setup for this testing is shown in Figure 41.

Figure 41 Dynamic Handle Force Testing

The method for dynamic handle testing is detailed below:

1. Load cell components attached to lifter
2. Volunteer height and weight and chair height measured
3. Lifter mechanism and configuration noted
4. Lift and test procedure discussed
5. Motion capture as discussed in Section 6.3, patient lifted and placed back on seat – control lift
6. Patient lifted to transit position and held for five seconds
7. Patient lowered onto seat and released from chest straps

The main application of this testing was to accurately assess handle forces present using the HTS3 mechanism. A dynamic assessment allowed analysis of the force profile providing a clearer understanding of the lift. Whether there was a decrease in force due to momentum was also of interest. For this testing, twelve patients of varied patient characteristics where lifted with a spread of patients as shown in Table 17. The results of this testing are discussed in Section 9.4.

Table 17 Spread of Patient Heights and Weights for HTS3 Testing

	$40kg \leq x < 55kg$	$55kg \leq x < 70kg$	$70kg \leq x < 80kg$	$80kg \leq x < 100kg$	$100kg \leq x < 120kg$
$1400mm \leq x < 1500mm$	✓				
$1500mm \leq x < 1650mm$	✓	✓	✓		
$1650mm \leq x < 1800mm$		✓	✓✓	✓	
$1800mm \leq x < 2000mm$		✓	✓	✓	✓

6.3 Motion Capture Trajectory Validation

Evaluation through motion capture has been completed to track trajectories of the patient and mechanism. The method for motion capture testing is detailed below:

1. Metre stick taped onto floor beside patient's chair, or reference geometry noted
2. Camera set up approximately 3 metres from lifter to provide a suitable side view of lift
3. Marker dots placed on elbow, shoulder, hip, knee, and ankle joints
4. Video recorded while patient lift occurs as discussed in Section 6.2.

6.4 Quasi-static Centre of Mass Analysis

Relevance of the location of a patient's centre of mass and the implication of the position on the patient and carer forces during a lift required assessment. Initially, this was based the values of de Leva, (1996). It is noted that adjustments due to ageing have been published (Haong & Mombaur, 2015). While it is possible to measure the patient's centre of mass position using a balance table, the lifted centre of mass is more difficult to assess (Virmavirta & Isolehto, 2014). Using a known lift trajectory with a known mechanism, the handle force for a patient of known height and weight can be measured. The vertical force on the footplate was also measured. Knowing these forces, the patient's LCM can be calculated by working backwards through the patient and lifter forces and discussed in Section 4.7.

The method for quasi-static centre of mass analysis is detailed below:

1. Load cell components attached to lifter
2. Scales attached to kneepad
3. Volunteer height and weight and chair height measured
4. Lifter mechanism and configuration noted
5. Lift and test procedure discussed
6. Patient fully lifted and placed back on seat – control lift
7. Patient lifted to the point where the thigh loses contact with chair
8. Patient lifted until α at 45 degrees and held steady at this point, ensuring no momentum present
9. Photo taken to calculate angles, footplate force and handle force measured at this point
10. Patient lifted until α at 60 degrees and held steady at this point, ensuring no momentum present

11. Photo taken to calculate angles, footplate force and handle force measured at this point
12. Patient lowered onto seat and released from any chest straps

It was found, in practice, that the LCM could be sufficiently estimated for non-bariatric patients using an adjustment factor relating the Body Mass Index of a patient. This numerical approximation is shown in Equation 37.

$$COM_{Adj} = 3.5\left(\frac{W}{H^2} - 23\right) \quad [37]$$

7 SINGLE PIVOT

7.1 CONCEPT

Within Section 5.1, it is seen that the single pivot position to approximate a zero force lift is located with Zones A and D. It was decided that, to minimise forces but still allow the pivot to be placed within the acceptable zone, the pivot position should be placed as far back towards the patient as possible without being located in Zone A.

7.2 LITTLE BLUE LIFTER

A previous lifter version, Little Blue, used a pivot placed at the extremity of Zone A at a height of 425 millimetres above the floor level as shown in Figure 42.

Figure 42 Little Blue Lifter

It was noted that the main limitations of the Little Blue mechanism were due to the geometry of the mechanism itself. The usability and comfort of the lifter was due to the inability to alter the pivot arm length. Functionality issues rose from the use of two pivots, at either side of the patient, resulting in collision with some lifting. The performance of the Single Pivot DSR is 42 and is shown in Table 18.

Table 18 Design Specifications, Little Blue Lifter Mechanism Performance Shown in Grey

<table>
<tr><th rowspan="2">Success Criteria / Value</th><th>Ease of Use</th><th>Carer Input</th><th>Safety</th><th>Stability</th><th>Cost</th><th>Manoeuvrability</th><th>Cognitive Requirement</th></tr>
<tr><th>Time taken for transfer (TT)</th><th>Carer force as percentage of body weight (CF)</th><th>Number of the following criteria met (MC)</th><th>Percentage of weight patient-supported (WB)</th><th>Estimated cost of device in NZD (DC)</th><th>Turning circle radius (TC)</th><th>Patient cognitive requirement (CR)</th></tr>
<tr><td>1</td><td>$10\ min < TT$</td><td>$80\% < CF$</td><td rowspan="10">• No carer forward back bending
• Load max 16kg
• Locking mechanism
• No twisting of carer torso
• No dragging of patient
• Force spread over large portion of patient's body
• Varied surface height
• No trap points
• Transportable
• Operation errors easily reversed</td><td>$100\% \leq WB$</td><td>$\$5000 \leq DC$</td><td>$2.2m \leq TC$</td><td rowspan="2">$100\%\ CR$</td></tr>
<tr><td>2</td><td>$8\ min < TT \leq 10\ min$</td><td>$70\% < CF \leq 80\%$</td><td>$80\% < WB < 100\%$</td><td>$\$4000 < DC < \$5000$</td><td>$2.1m < TC < 2.2m$</td></tr>
<tr><td>3</td><td>$6\ min < TT \leq 8\ min$</td><td>$60\% < CF \leq 70\%$</td><td>$70\% < WB \leq 80\%$</td><td>$\$3000 < DC \leq \$4000$</td><td>$2.0m < TC \leq 2.1m$</td><td rowspan="2">Able to stay alert and focused for entire lift</td></tr>
<tr><td>4</td><td>$5\ min < TT \leq 6\ min$</td><td>$50\% < CF \leq 60\%$</td><td>$60\% < WB \leq 70\%$</td><td>$\$2000 < DC \leq \$3000$</td><td>$1.8m < TC \leq 2.0m$</td></tr>
<tr><td>5</td><td>$4\ min < TT \leq 5\ min$</td><td>$40\% < CF \leq 50\%$</td><td>$50\% < WB \leq 60\%$</td><td>$\$1000 < DC \leq \$2000$</td><td>$1.6m < TC \leq 1.8m$</td><td rowspan="2">Able to follow a series of simple instructions</td></tr>
<tr><td>6</td><td>$3\ min < TT \leq 4\ min$</td><td>$30\% < CF \leq 40\%$</td><td>$\mathbf{30\% < WB \leq 50\%}$</td><td>$\mathbf{\$750 < DC \leq \$1000}$</td><td>$\mathbf{1.4m < TC \leq 1.6m}$</td></tr>
<tr><td>7</td><td>$2\ min < TT \leq 3\ min$</td><td>$20\% < CF \leq 30\%$</td><td>$20\% < WB \leq 30\%$</td><td>$\$500 < DC \leq \$750$</td><td>$1.2m < TC \leq 1.4m$</td><td rowspan="2">Able to follow a simple instruction</td></tr>
<tr><td>8</td><td>$\mathbf{1\ min < TT \leq 2\ min}$</td><td>$\mathbf{10\% < CF \leq 20\%}$</td><td>$10\% < WB \leq 20\%$</td><td>$\$250 < DC \leq \$500$</td><td>$1.0m < TC \leq 1.2m$</td></tr>
<tr><td>9</td><td>$30\ s < TT \leq 1\ min$</td><td>$5\% < CF \leq 10\%$</td><td>$0\% < WB \leq 10\%$</td><td>$\$100 < DC \leq \$250$</td><td>$0.5m < TC \leq 1.0\ m$</td><td rowspan="2">$0\%\ CR$</td></tr>
<tr><td>10</td><td>$TT \leq 30\ s$</td><td>$CF \leq 5\%$</td><td>$WB \leq 0\%$</td><td>$DC \leq \$100$</td><td>$TC \leq 0.5m$</td></tr>
</table>

7.3 Final Geometry

Length of the pivot arm reduced the functionality of the Little Blue lifter. Therefore, it was decided that an upgraded single pivot mechanism would be used, with the ability to adjust the length of the pivot arm. The pivot was placed centrally, vertically above the ankle joint. The transport position was increased by extended the trajectory of the lift to the point where the torso angle, α, was equal to 70 degrees. This single pivot lifter is shown in Figure 43.

Figure 43 Single Pivot Lifter

It should be noted that the pivot height was taken from the footplate rather than the floor level due to the uncertainty of floor level on carpets and compliant or impact-absorbing flooring. The footplate is located 50 millimetres above floor level on a flat, concrete floor. The geometry of the Single Pivot Lifter is shown in Figure 44.

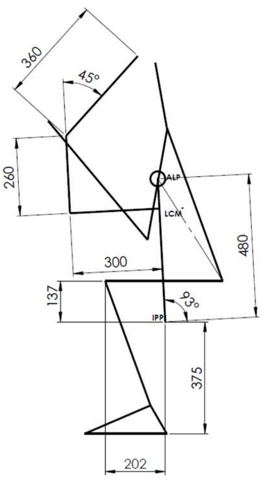


Figure 44 Geometry and Dimensions of Single Pivot Lifter

It must be noted that the 93˚ angle noted is included only to identify the initial position of the mechanism for the sample patient. This will increase to 137˚ at the transport position, where α is 70˚ and β is 33˚. It is expected that these angles will differ with respect to patient height.

7.4 Coding and Simulation

Key to the simulation is the Matlab code used to generate results for the single pivot shown in Appendix D; a flow chart of the iteration process is included in Figure 45.

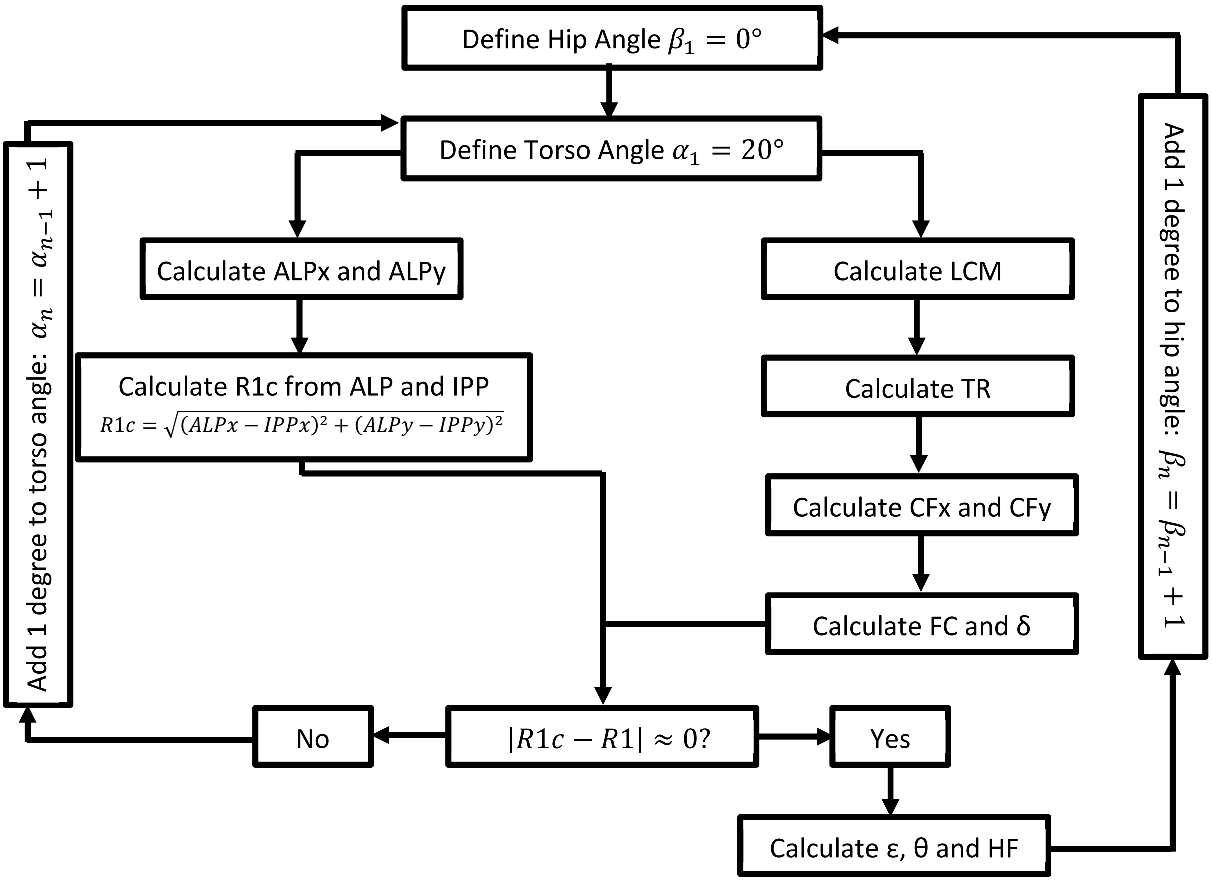


Figure 45 Flowchart of Iteration Process for Single Pivot Lifter Simulation

Before this code was relied on to assess the carer input and mechanism forces the model was validated to ensure that the calculations and simulations suitably emulated the measured lift forces.

7.5 Testing and Results

Assessment of the motion capture footage was completed to validate the simulated trajectories for a single pivot lifter. The motion capture and simulated trajectories were the compared as shown in Figure 46.

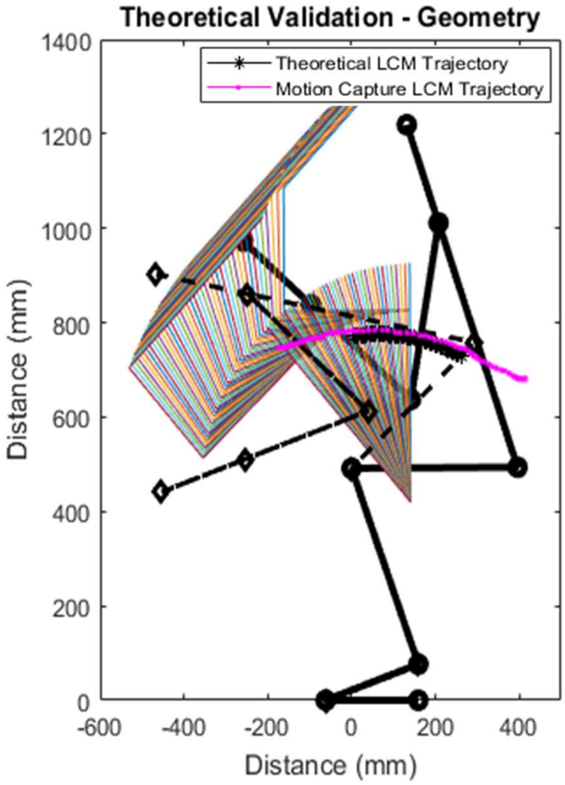


Figure 46 Single Pivot Trajectory Comparison and Validation

It is seen that the motion capture of the LCM position closely follows the theoretical path although the trajectory extends further than the theoretical trajectory. The height range of patients is limited to between 1400 and 2100 millimetres. To validate the handle force calculations, three patients of known height and weight were lifted with the standard chest pad fitted. The patient characteristics are shown in Table 19; R1T and R1P indicate the theoretical and practical values of R1 respectively.

Table 19 Patient Characteristics for Single Pivot Validation

Patient	Height (mm)	Weight (kg)	Scale Factor	R1T (mm)	R1P (mm)
P1	1580	57	0.88	422	400
P2	1800	82	1.01	485	450
P3	1830	76	1.02	490	450

The handle force along the ALP trajectory was then measured at three points along the lift using a spring balance. The torso angle from the vertical position was also measured at each point as shown in Table 20.

Table 20 Measured Handle Forces for Single Pivot Mechanism Testing

Force	Person	Handle Force at Torso Angle			
		20˚	45˚	50˚	60˚
Measured Force (N)	1	50	30	5	0
	2	130	80	-	30
	3	100	110	-	40
Calculated Force (N)	1	52	26	17	-1
	2	132	75	57	16
	3	120	120	81	56
Measured with 40mm kneepad (N)	1	60	30	-	0
	2	130	90	-	50
	3	110	120	80	-
Calculated with 40mm kneepad (N)	1	62	25	36	-4
	2	132	96	83	46
	3	118	115.2	108	81.5

These points were then plotted against the simulated force profile as shown in Figure 47.

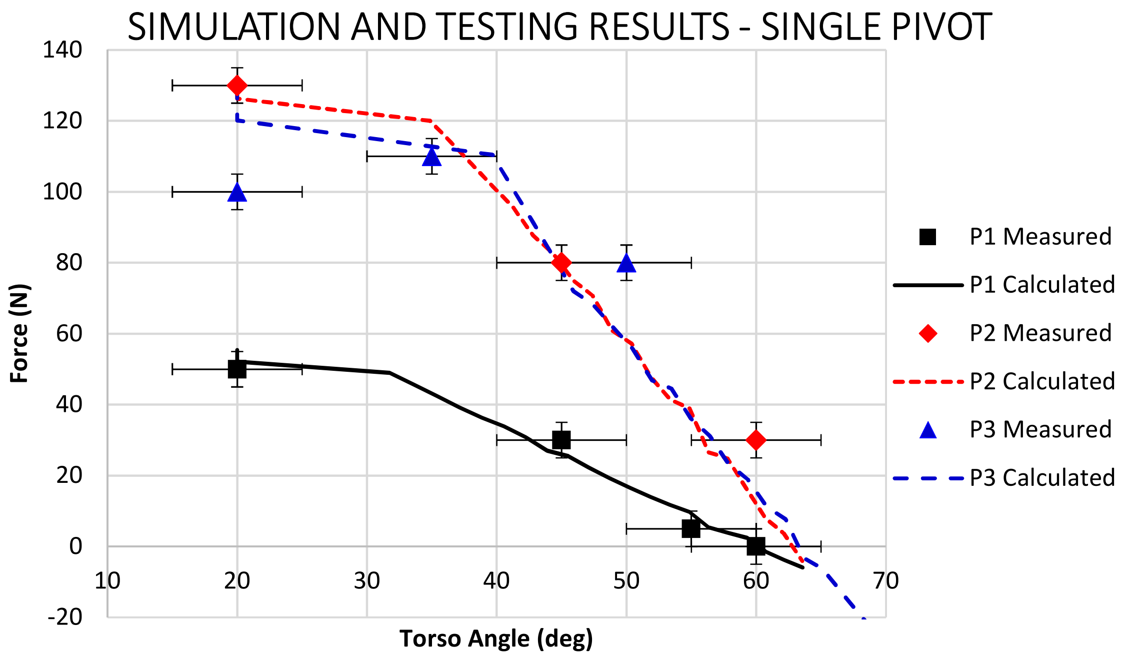


Figure 47 Simulation and Testing Result Comparison for Standard Single Pivot Mechanism

It is seen that the calculated and measured results for Person 1 and Person 2 closely match, with most data points within range of the calculated forces. It is expected that the initial torso angle for Person 3 was not accurately measured. It appears the correct torso angle may be 5 degrees less than measured. This would result in the patient not being fully lifted off the seat at the 20 degree torso angle measurement and, as would be expected, the handle force is lower than anticipated. Testing was then completed with 40 millimetre blocks on the kneepads simulating the pivot point being shifted 40 millimetres horizontally towards the knees. The results of this test are shown in Figure 48.

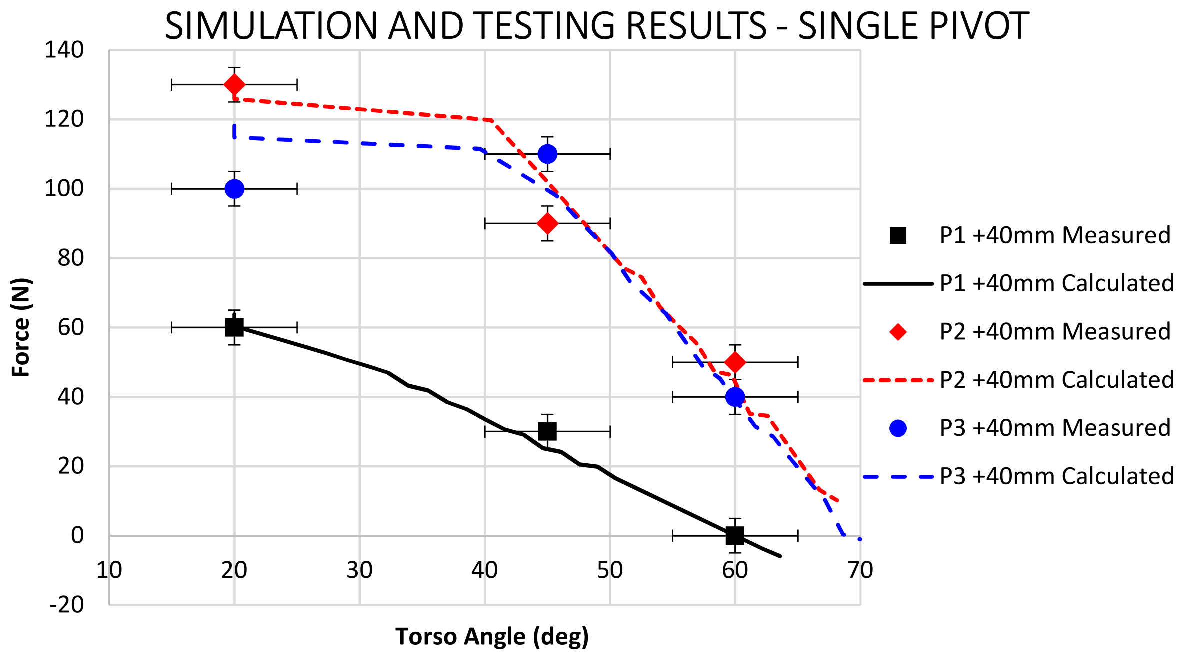


Figure 48 Simulation and Testing Result Comparison with 40 millimetre Knee Pad Adjustment

It can be seen that the calculated and measured results for Person 1 and Person 2 more closely match during this test, with all measured data points solidly within range of the calculated forces.

7.6 Findings

Little Blue simulation results are shown in Table 21.

Table 21 Maximum Handle Forces (Newtons) for Varied Patient Heights and Weights with R1 Constant for All Heights

		Height (mm)												
		1400	**1450**	**1500**	**1550**	**1600**	**1650**	**1700**	**1750**	**1800**	**1850**	**1900**	**1950**	**2000**
Weight (kg)	**40**	57.5	56.7	56.2	56.1	56.4	57.0	58.0	59.4	61.1	63.3	65.8	68.6	71.8
	45	65.7	64.6	63.8	63.5	63.6	64.1	65.0	66.3	68.1	70.3	72.9	75.9	79.3
	50	74.0	72.5	71.5	70.9	70.8	71.2	72.0	73.3	75.1	77.3	80.0	83.1	86.7
	55	82.2	80.4	79.1	78.3	78.0	78.2	79.0	80.2	82.0	84.3	87.1	90.4	94.2
	60	90.4	88.2	86.7	85.7	85.2	85.3	86.0	87.2	89.0	91.3	94.2	97.6	101.6
	65	98.6	96.1	94.3	93.1	92.4	92.4	93.0	94.1	95.9	98.3	101.3	104.9	109.0
	70	106.8	104.0	101.9	100.4	99.6	99.5	100.0	101.1	102.9	105.3	108.4	112.1	116.5
	75	115.0	111.9	109.5	107.8	106.8	106.6	106.9	108.0	109.8	112.3	115.5	119.4	123.9
	80	123.2	119.8	117.1	115.2	114.1	113.6	113.9	115.0	116.8	119.3	122.6	126.6	131.4
	85	131.4	127.7	124.7	122.6	121.3	120.7	120.9	121.9	123.7	126.3	129.7	133.9	138.8
	90	139.6	135.6	132.4	130.0	128.5	127.8	127.9	128.9	130.7	133.3	136.8	141.1	146.2
	95	147.8	143.4	140.0	137.4	135.7	134.9	134.9	135.8	137.6	140.3	143.9	148.4	153.7
	100	156.0	151.3	147.6	144.8	142.9	141.9	141.9	142.8	144.6	147.3	151.0	155.6	**161.1**

These results are expected to be the same as the Little Blue lifter, as R1 does not alter with patient height. It should be noted that 157 Newtons is only exceeded for the case where a patient has a height and weight of 2000 millimetres and 100 kilograms respectively. As mentioned in Section 7.2, there are comfort and functionality issues that occur when R1 remains constant. Considering this, the simulation was completed with R1 varying by a scale factor as shown in Equation 38.

$$SF = \frac{H}{1800} \quad [38]$$

The results of simulation with varied pivot length for a variety of patients with weights and heights are shown in Table 22.

Table 22 Maximum Handle Force (Newtons) for Varied Patient Heights and Weights with Adjusted Pivot Length

		Height (mm)												
		1400	**1450**	**1500**	**1550**	**1600**	**1650**	**1700**	**1750**	**1800**	**1850**	**1900**	**1950**	**2000**
Weight (kg)	**40**	34.3	37.5	40.6	43.8	47.0	50.2	53.4	56.5	59.7	62.9	66.1	69.3	72.4
	45	36.9	40.6	44.3	48.0	51.7	55.3	59.0	62.7	66.4	70.1	73.7	77.4	81.1
	50	39.6	43.8	48.0	52.1	56.3	60.5	64.7	68.9	73.0	77.2	81.4	85.6	89.8
	55	42.3	46.9	51.6	56.3	61.0	65.7	70.3	75.0	79.7	84.4	89.1	93.7	98.4
	60	44.9	50.1	55.3	60.5	65.6	70.8	76.0	81.2	86.4	91.5	96.7	101.9	107.1
	65	47.6	53.3	58.9	64.6	70.3	76.0	81.7	87.3	93.0	98.7	104.4	110.1	115.7
	70	50.2	56.4	62.6	68.8	74.9	81.1	87.3	93.5	99.7	105.8	112.0	118.2	124.4
	75	52.9	59.6	66.2	72.9	79.6	86.3	93.0	99.6	106.3	113.0	119.7	126.4	133.0
	80	55.5	62.7	69.9	77.1	84.3	91.4	98.6	105.8	113.0	120.2	127.3	134.5	141.7
	85	58.2	65.9	73.6	81.2	88.9	96.6	104.3	112.0	119.6	127.3	135.0	142.7	150.4
	90	60.9	69.0	77.2	85.4	93.6	101.8	109.9	118.1	126.3	134.5	142.7	150.8	**159.0**
	95	63.5	72.2	80.9	89.6	98.2	106.9	115.6	124.3	133.0	141.6	150.3	**159.0**	**167.7**
	100	66.2	75.4	84.5	93.7	102.9	112.1	121.3	130.4	139.6	148.8	**158.0**	**167.2**	**176.3**

As well as forces increasing with weight, the handle forces increase as the height of the patient increases. It is expected that this is due to the resultant chest pad force angle, δ, increasing as the height of the patient increases. It should be noted that the 16 kilogram limit is exceeded at lower heights and weights than when R1 is constant. This is due to the LCM of taller patients being lifted higher as R1 is increased. Although this causes an increase in handle force, it is recommended that R1 is increased for taller patients as this allows a larger hip angle, allowing a more comfortable lift.

7.7 Summary

Obviously while the Single Pivot concept addressed the main design issues of the Little Blue Lifter, functionality issues were still present. The main issue with the single pivot mechanism was found to be the location of the pivot. This was set to be vertically above the ankle joint. However, in practice, it was found that throughout the lift the patient's knee sunk further into the foam of the kneepads. This resulted in the angle of the lower leg increasing, causing the mechanism to collide with the chair before the patient's knees were completely in contact with the kneepads, and placing the patient further towards the front of that chair than intended.

The single pivot concept is only compatible with passive chest pads, as the patient is required to rotate around the chest pad throughout the lift. This limits the mechanism to be used only by patients with suitable upper body strength and balance. Active chest pads that held the patient securely in place would allow the patient to be lifted with negligible patient input and would stop any overbalancing or instability. Active chest pad may also increase patient comfort, as the force applied during the lift will be placed over a larger surface area of the body.

The handle forces for the carer are at the upper extreme of the allowable force band. From this, it is taken that the LCM cannot be raised any higher, or the ALP trajectory made any steeper without effecting the handle forces and raising these to higher than the allowable force band. The performance of the Single Pivot DSR is 43 and is shown in Table 23.

Table 23 Design Specifications, Single Pivot Lifter Mechanism Performance Shown in Grey

Value \ Success Criteria	**Ease of Use**	**Carer Input**	**Safety**	**Stability**	**Cost**	**Manoeuvrability**	**Cognitive Requirement**
	Time taken for transfer (TT)	Carer force as percentage of body weight (CF)	Number of the following criteria met (MC)	Percentage of weight patient-supported (WB)	Estimated cost of device in NZD (DC)	Turning circle radius (TC)	Patient cognitive requirement (CR)
1	$10\ min < TT$	$80\% < CF$	• No carer forward back bending	$100\% \leq WB$	$\$5000 \leq DC$	$2.2m \leq TC$	$100\%\ CR$
2	$8\ min < TT \leq 10\ min$	$70\% < CF \leq 80\%$	**• Load max 16kg**	$80\% < WB < 100\%$	$\$4000 < DC < \$5000$	$2.1m < TC < 2.2m$	
3	$6\ min < TT \leq 8\ min$	$60\% < CF \leq 70\%$	• Locking mechanism	$70\% < WB \leq 80\%$	$\$3000 < DC \leq \$4000$	$2.0m < TC \leq 2.1m$	**Able to stay alert and focused for entire lift**
4	$5\ min < TT \leq 6\ min$	$50\% < CF \leq 60\%$	**• No twisting of carer torso**	$60\% < WB \leq 70\%$	$\$2000 < DC \leq \$3000$	$1.8m < TC \leq 2.0m$	
5	$4\ min < TT \leq 5\ min$	$40\% < CF \leq 50\%$	• No dragging of patient	$50\% < WB \leq 60\%$	$\$1000 < DC \leq \$2000$	$1.6m < TC \leq 1.8m$	Able to follow a series of simple instructions
6	$3\ min < TT \leq 4\ min$	$30\% < CF \leq 40\%$	• Force spread over large portion of patient's body	$\mathbf{30\% < WB \leq 50\%}$	$\$750 < DC \leq \$1000$	$\mathbf{1.4m < TC \leq 1.6m}$	
7	$\mathbf{2\ min < TT \leq 3\ min}$	$20\% < CF \leq 30\%$	**• Varied surface height**	$20\% < WB \leq 30\%$	$\mathbf{\$500 < DC \leq \$750}$	$1.2m < TC \leq 1.4m$	Able to follow a simple instruction
8	$1\ min < TT \leq 2\ min$	$\mathbf{10\% < CF \leq 20\%}$	**• No trap points**	$10\% < WB \leq 20\%$	$\$250 < DC \leq \$500$	$1.0m < TC \leq 1.2m$	
9	$30\ s < TT \leq 1\ min$	$5\% < CF \leq 10\%$	**• Transportable**	$0\% < WB \leq 10\%$	$\$100 < DC \leq \$250$	$0.5m < TC \leq 1.0\ m$	$0\%\ CR$
10	$TT \leq 30\ s$	$CF \leq 5\%$	**• Operation errors easily reversed**	$WB \leq 0\%$	$DC \leq \$100$	$TC \leq 0.5m$	

It was decided that the mechanism must be adapted to allow the use of an active chest pad. It was anticipated that an active chest pad would support the patient throughout the lift and help the patient move forwards from the back of a seat. It was also decided to move the pivot further from Zone A to ensure there were no collision issues. The mechanism chosen to develop was a double pivot mechanism and is discussed in Section 9.

8 HTS2 Tilting Chest Pad

8.1 Concept

Noted in Section 7.7 are the issues present with a single pivot mechanism. To address these issues, a Double Pivot or Tilting Chest Pad lifter was developed. The concept used two pivot points to follow an approximation of the Little Blue and Single Pivot trajectory while ensuring the pivot point was set further from Zone A than either the Little Blue or Single Pivot mechanisms. It was decided that, as the secondary pivot arm would move with the patient, the pivot location could enter Zone C.

The secondary pivot arm is constrained as discussed previously in Section 5.1, with the pivot to be located outside Zones A, B, C, D, and E. The IPP is located 45 millimetres above the footplate level, or 95 millimetres from the floor level and 130 millimetres nearer the patient from the knee joint.

The pivot arm lengths are 623 millimetres and 120 millimetres for the lower and upper pivot arms respectively. The Tilting Chest Pad concept is shown in Figure 49.

Figure 49 Tilting Chest Pad Lifter HTS2

8.2 Final Geometry

Expected height of the footplate is located 50 millimetres above floor level on a flat, concrete floor. It should be noted that the pivot height was taken from the footplate rather than the floor level due to the uncertainty of floor level on carpets and compliant or impact-absorbing flooring. The geometry of the tilting chest pad mechanism is shown in Figure 50.

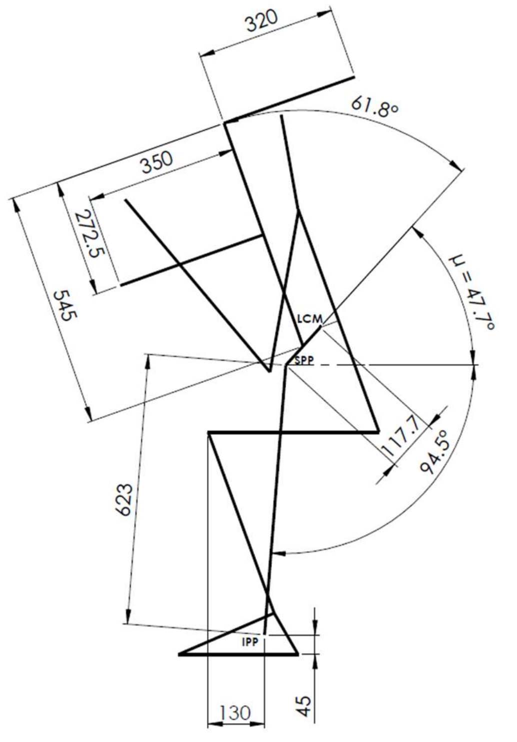


Figure 50 Geometry and Dimensions of Tilting Chest Pad Mechanism

It must be noted that the angles noted are included only to identify the initial position of the mechanism for the sample patient. It is expected that these angles will differ with respect to patient height.

8.3 Coding and Simulation

It is important to note that the ALP position defined as the point around which the patient rotates. For the case of the tilting chest pad, the ALP position is located at the SPP. This results in the ALP being calculated and handled slightly differently within the code. To simplify the handling of this, the Virtual Load Point or VLP is introduced. For the single pivot case, the ALP and VLP are located at the same position. The VLP is defined as the torso's point of contact with the chest pad. For active chest pad cases where the torso is held rigidly, the VLP can be approximated as being located at the LCM. As the torso will follow through the same angle as the VLP, the ALP position can be calculated using the equations 39 and 40.

$$ALPx = VLPx - R2\cos(\alpha + \mu) \quad [39]$$

$$ALPy = VLPy - R2\sin(\alpha + \mu) \quad [40]$$

The length of R1 from the calculated ALP position can be assessed against the known value of R1, similarly to the single pivot calculations. The Matlab code used to generate results for the tilting chest pad mechanism is shown in Appendix G; a flow chart of the iteration process is included in Figure 51.

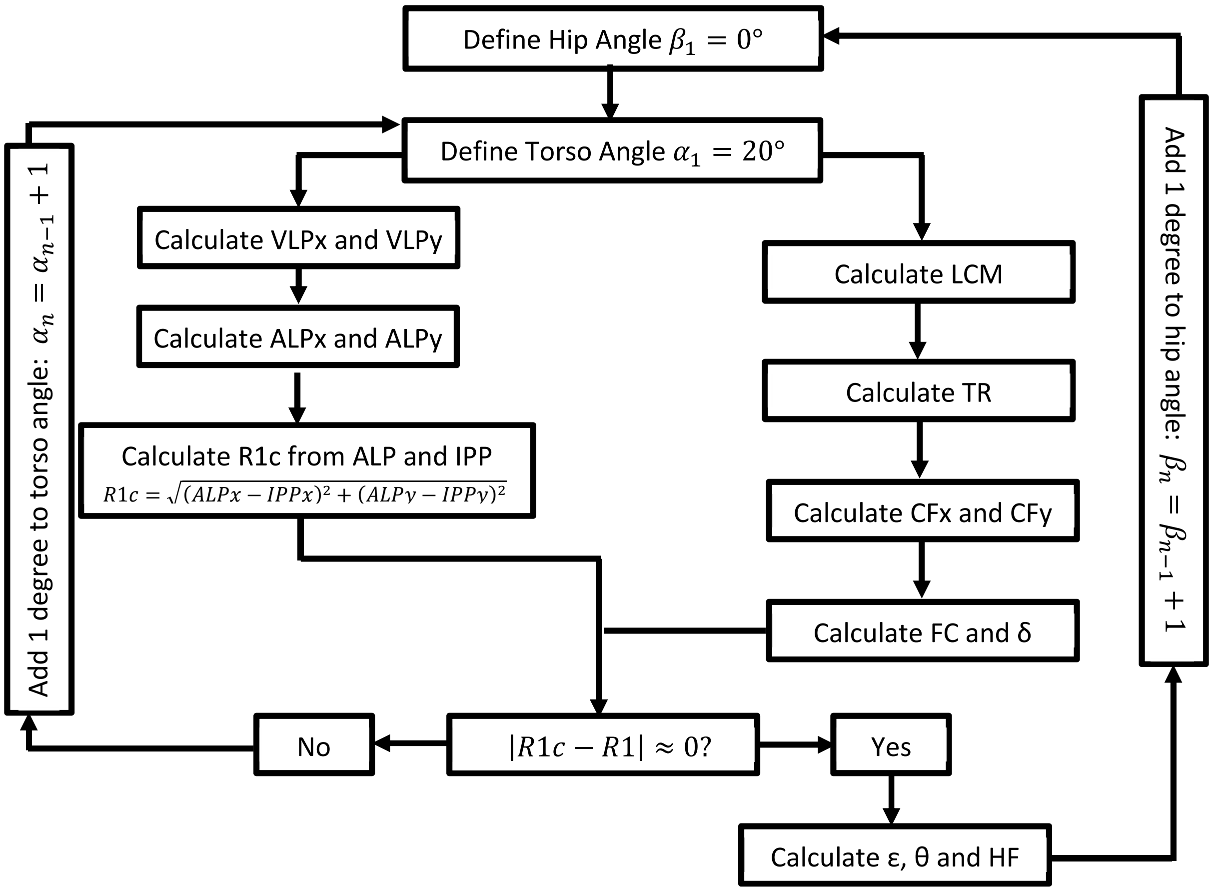


Figure 51 Flowchart of Iteration Process for Tilting Chest Pad Lifter Simulation

8.4 TESTING AND RESULTS

To validate that the simulated trajectories, motion capture was used to assess video footage of the single pivot lifter. The motion capture and simulated trajectories were the compared as shown in Figure 52 .

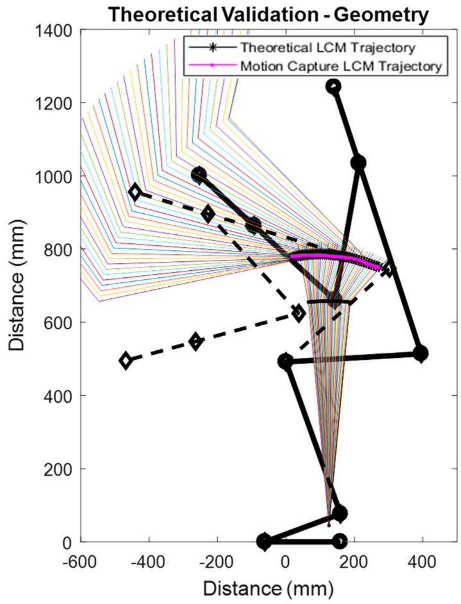


Figure 52 HTS2 Tilting Chest Pad Trajectory Comparison and Validation

To validate the handle force calculations three patients of known height and weight, as shown in Table 24, were lifted.

Table 24 Patient Characteristics for HTS2 Validation

Patient	Height (mm)	Weight (kg)
P1	1570	50
P2	1800	82
P3	1770	75

The secondary pivot arm length and chosen chest pad were noted. The handle force along the ALP trajectory was then measured at three points along the lift using a spring balance. The torso angle from the vertical position was also measured at each point as shown in Table 27.

Table 25 Measured Handle Forces for HTS2 Mechanism Testing

Force	Person	Handle Force at Torso Angle		
		25˚	**45˚**	**60˚**
Measured Force (N)	1	50	20	5
	2	80	40	0
	3	75	25	0
Calculated Force (N)	1	47.0	15.6	-0.2
	2	81.4	23.8	-8.6
	3	72.2	24.0	-5.1

The initial torso angle measurement was completed at 25 degrees rather than 20 degrees as this eliminated the possibility that the patient's weight had not been completely removed from the chair. These points were then plotted against the simulated force profile as shown in Figure 53.

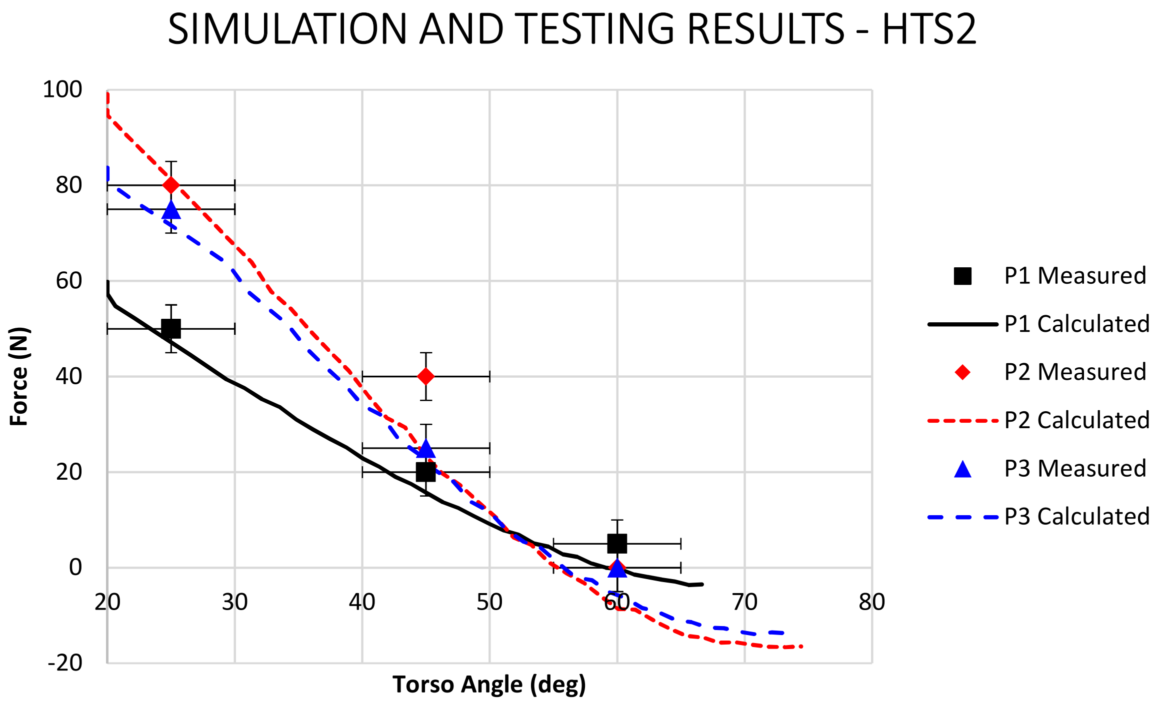


Figure 53 Simulation and Testing Result Comparison for HTS2

It is seen that the calculated and measured results closely match, with most measured data points within range of the calculated forces. Uncertainties that arise from testing are shown using error bars of ± 5 Newtons and ± 5 degrees.

This was also validated through the use of motion capture and dynamic handle force testing with Patient 1 to verify the testing techniques. The results of testing are shown in Figure 54.

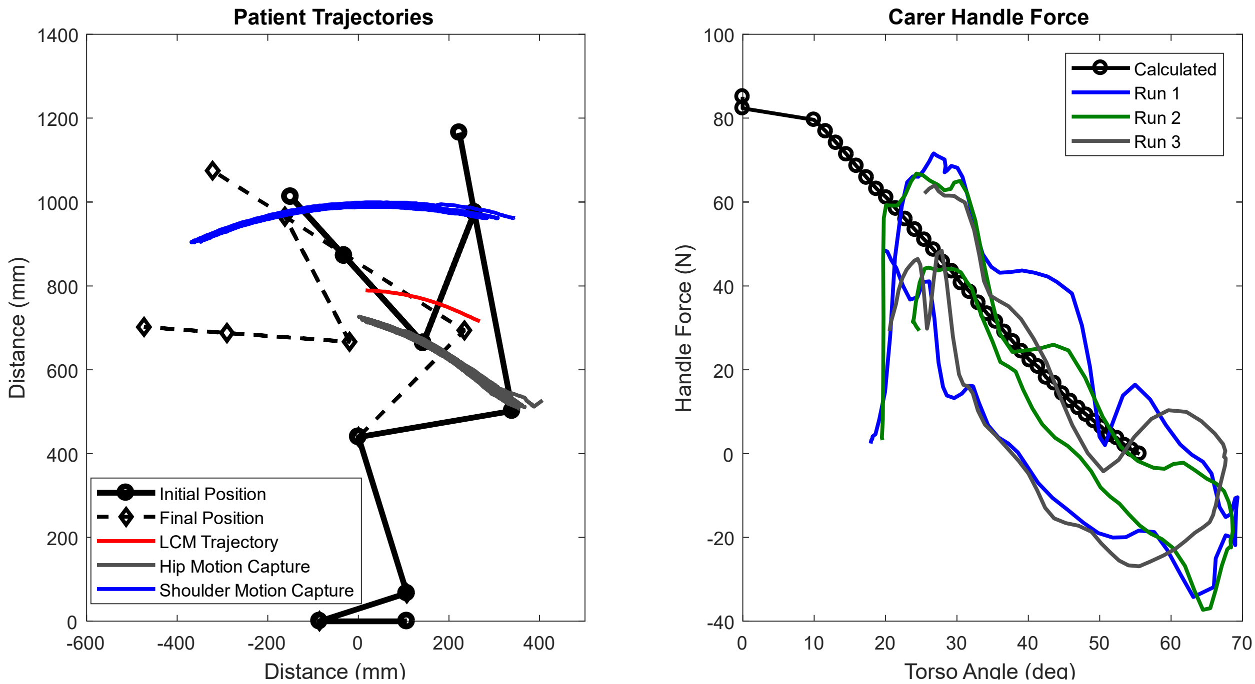


Figure 54 HTS2 Testing Through Dynamic Handle Force Testing

From this, it can be seen there are some interesting differences between the calculated and measured forces and trajectories. When observing the patient trajectories it is noticed that the hip is not raised as high in reality as the simulation suggests. It is expected that this is due to the rigidity of the model not allowing for any bending of the body segments. The bending of the back during the lift results in the hip to thigh angle becoming more acute than expected, further increasing the patient discomfort. It should also be noted that the measured peak force is lower than the anticipated handle force. It is expected that this is due to the patient's weight being supported gradually by the mechanism, with some of the patient's weight being supported by the chair or transfer surface.

8.5 Summary

While the Tilting Chest Pad concept addressed the main design issues of the Single Pivot, functionality issues were still present. The main issue with the tilting chest pad mechanism was found to be that the hip angle was too acute during the lift, and especially in the transport position. This was found to be uncomfortable and not feasible for low mobility patients. This angle was found to be as low as 39.9 degrees for a 2000 millimetre patient, as shown in Figure 55 .

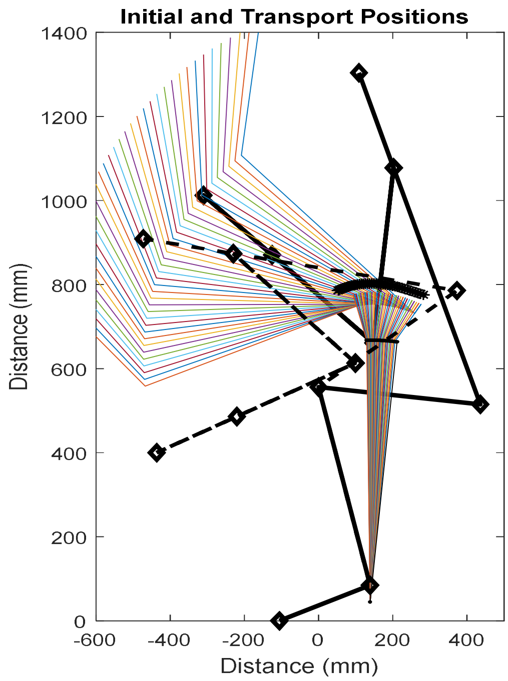


Figure 55 Initial and Transport Positions for Tilting Chest Pad Mechanism for a 2000mm Patient

The acuteness in the hip arises from the need to keep forces reduced to below 157 Newtons throughout the lift. To ensure this maximum force is not exceeded, the LCM of the patient can only be raised gradually when commencing the lift. Due to the geometry restraints, the LCM is then lowered towards the end of the lift, decreasing the angle between the thigh and torso.

The height range of patients using this mechanism is limited to between 1400 and 2100 millimetres. The handle forces for the carer are at the upper extreme of the allowable force band. From this, it is taken that the LCM cannot be raised any higher, or the ALP trajectory made any steeper without effecting the handle forces and raising these to higher than the allowable force band. The performance of the Tilting Chest Pad mechanism DSR is 45 and is shown in Table 26 .

Table 26 Design Specifications, HTS2 Lifter Mechanism Performance Shown in Grey

<table>
<tr><th rowspan="2">Success Criteria / Value</th><th>Ease of Use</th><th>Carer Input</th><th>Safety</th><th>Stability</th><th>Cost</th><th>Manoeuvrability</th><th>Cognitive Requirement</th></tr>
<tr><td>Time taken for transfer (TT)</td><td>Carer force as percentage of body weight (CF)</td><td>Number of the following criteria met (MC)</td><td>Percentage of weight patient-supported (WB)</td><td>Estimated cost of device in NZD (DC)</td><td>Turning circle radius (TC)</td><td>Patient cognitive requirement (CR)</td></tr>
<tr><td>1</td><td>$10\ min < TT$</td><td>$80\% < CF$</td><td rowspan="10">• No carer forward back bending
• Load max 16kg
• Locking mechanism
• No twisting of carer torso
• No dragging of patient
• Force spread over large portion of patient's body
• Varied surface height
• No trap points
• Transportable
• Operation errors easily reversed</td><td>$100\% \leq WB$</td><td>$\$5000 \leq DC$</td><td>$2.2m \leq TC$</td><td rowspan="2">$100\%\ CR$</td></tr>
<tr><td>2</td><td>$8\ min < TT \leq 10\ min$</td><td>$70\% < CF \leq 80\%$</td><td>$80\% < WB < 100\%$</td><td>$\$4000 < DC < \$5000$</td><td>$2.1m < TC < 2.2m$</td></tr>
<tr><td>3</td><td>$6\ min < TT \leq 8\ min$</td><td>$60\% < CF \leq 70\%$</td><td>$70\% < WB \leq 80\%$</td><td>$\$3000 < DC \leq \$4000$</td><td>$2.0m < TC \leq 2.1m$</td><td rowspan="2">Able to stay alert and focused for entire lift</td></tr>
<tr><td>4</td><td>$5\ min < TT \leq 6\ min$</td><td>$50\% < CF \leq 60\%$</td><td>$60\% < WB \leq 70\%$</td><td>$\$2000 < DC \leq \$3000$</td><td>$1.8m < TC \leq 2.0m$</td></tr>
<tr><td>5</td><td>$4\ min < TT \leq 5\ min$</td><td>$40\% < CF \leq 50\%$</td><td>$50\% < WB \leq 60\%$</td><td>$\mathbf{\$1000 < DC \leq \$2000}$</td><td>$1.6m < TC \leq 1.8m$</td><td rowspan="2">Able to follow a series of simple instructions</td></tr>
<tr><td>6</td><td>$\mathbf{3\ min < TT \leq 4\ min}$</td><td>$30\% < CF \leq 40\%$</td><td>$\mathbf{30\% < WB \leq 50\%}$</td><td>$\$750 < DC \leq \$1000$</td><td>$\mathbf{1.4m < TC \leq 1.6m}$</td></tr>
<tr><td>7</td><td>$2\ min < TT \leq 3\ min$</td><td>$20\% < CF \leq 30\%$</td><td>$20\% < WB \leq 30\%$</td><td>$\$500 < DC \leq \$750$</td><td>$1.2m < TC \leq 1.4m$</td><td rowspan="2">Able to follow a simple instruction</td></tr>
<tr><td>8</td><td>$1\ min < TT \leq 2\ min$</td><td>$\mathbf{10\% < CF \leq 20\%}$</td><td>$10\% < WB \leq 20\%$</td><td>$\$250 < DC \leq \$500$</td><td>$1.0m < TC \leq 1.2m$</td></tr>
<tr><td>9</td><td>$30\ s < TT \leq 1\ min$</td><td>$5\% < CF \leq 10\%$</td><td>$0\% < WB \leq 10\%$</td><td>$\$100 < DC \leq \$250$</td><td>$0.5m < TC \leq 1.0\ m$</td><td rowspan="2">$0\%\ CR$</td></tr>
<tr><td>10</td><td>$TT \leq 30\ s$</td><td>$CF \leq 5\%$</td><td>$WB \leq 0\%$</td><td>$DC \leq \$100$</td><td>$TC \leq 0.5m$</td></tr>
</table>

It was decided that the geometry of the mechanism must be adapted to allow for a greater vertical increase of LCM position throughout the lift. In addition, it was found there was a large variance in torso angles between patients when lifted; this is mainly due to variance in patient height. Due to this finding, it was decided that the most comfortable, effective, and simple active chest pad consists of a padded front with an adjustable strap which, when tightened, draws the patient towards the chest pad for support. It was decided to attach the kneepads to the pivot arm of the adjusted mechanism allowing the patient's shank angle to increase throughout the lift. This removes some of the calf strain that was present in the previous lifters and ensures the patient's heel maintains contact with the footplate at all times. This will also allow a patient to be retrieved from further back in a chair. Slowly increasing the angle of the shank may also aid in rehabilitation. The mechanism chosen to develop was an adjusted double pivot mechanism and is discussed in Section 9.

9 HTS3 Adjusted Tilting Chest Pad

9.1 Concept

Sections 7.7 and 8.5 discussed the issues present with single pivot and tilting chest pad mechanisms. While it was found that the HTS2 mechanism solved the design issues that arose with Single Pivot Mechanism, alterations to the design were required. In essence, the HTS3 design is an alteration of the HTS2 mechanism. The primary pivot arm is constrained as discussed previously in Section 5.1, with the pivot to be located outside Zones A, B, C, D, and E. The position of the IPP has not been altered from its location in the tilting chest pad design.

As discussed in Section 8.5, previous mechanisms have not accounted for cases where the shank is vertical. This greatly limits the functionality of these mechanisms, as patients sitting in easy chairs would need to shuffle forwards before the mechanism could lift them. Previous mechanisms also replace the patient near the front of seats, requiring them to reposition after the lift is completed. Moving kneepads allow the patient to be wheeled very close to a chair and allow them to be sat much nearer the back, promoting an upright sitting position. The moving kneepads also allow much more clearance behind the patient when seating them. A comparison of thigh collision and patient position on a 500 millimetre seat for the Little Blue and HTS3 is shown in Figure 56.

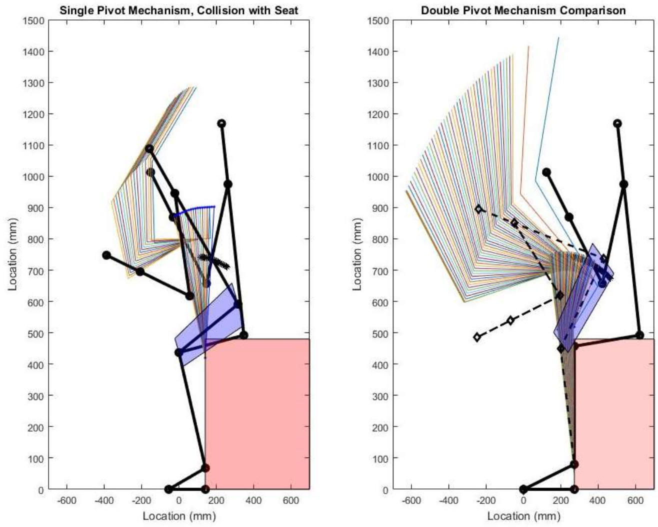


Figure 56 Transport Position for a 1600 Millimetre Tall Patient using Little Blue (Left) and HTS3 Lifters (Right)

It can be seen that, for the Little Blue lift case, the pivot arm length results in the patient's thighs colliding with the chair when being transporting back to their seat. There is also collision of the pivot point with the chair, resulting in the patient being placed closer to the edge of the seat. It is expected that the patient will then need be repositioned in the chair by carers. In comparison, the HTS3 allows a patient to be placed further back. The pivot point also causes the hip to be raised higher in the transport position to ensure clearance for a range of seat heights.

9.2 Final Geometry

The concept used two pivot points to lift the centre of mass of the patient approximately 80 millimetres higher than previos mechanisms. The lower pivot point at the same position as the HTS2 mechanism. The pivot arm lengths are 474 millimetres and 325 millimetres for the lower and upper pivot arms respectively. The HTS3 concept is shown in Figure 57 . The geometry of the HTS3 Lifter is shown in Figure 58 .

Figure 57 Adjusted Tilting Chest Pad HTS3

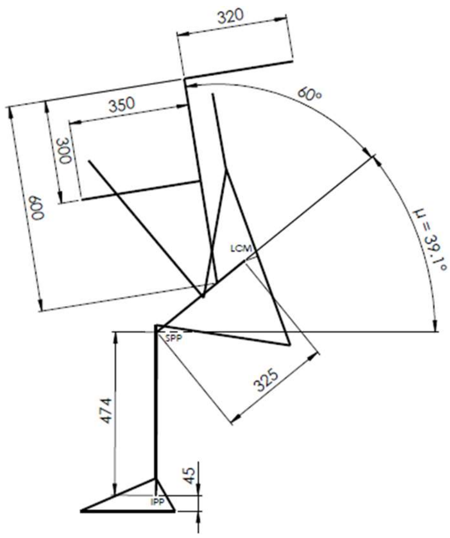


Figure 58 Geometry and Dimensions of HTS3 Mechanism

It must be noted that the initial position of the primary pivot arm mechanism is 90˚ from the positive horizontal direction for the sample patient. It is anticipated that this will increase to 106˚ at the transport position, where α is 56˚ and β is 37˚. These angles will differ with respect to patient height.

9.3 Coding and Simulation

The Matlab code used to generate results for the HTS3 is shown in Appendix H. Before this code was relied on to assess the carer input and mechanism forces the model was validated to ensure that the calculations and simulations suitably emulated the measured lift forces.

A numerical approximation of the expected peak force was developed to estimate the expected peak force for a patient of a defined height and weight. The approximation uses a polynomial fit to estimate the maximum handle force, a graphical representation of this is shown in Figure 59.

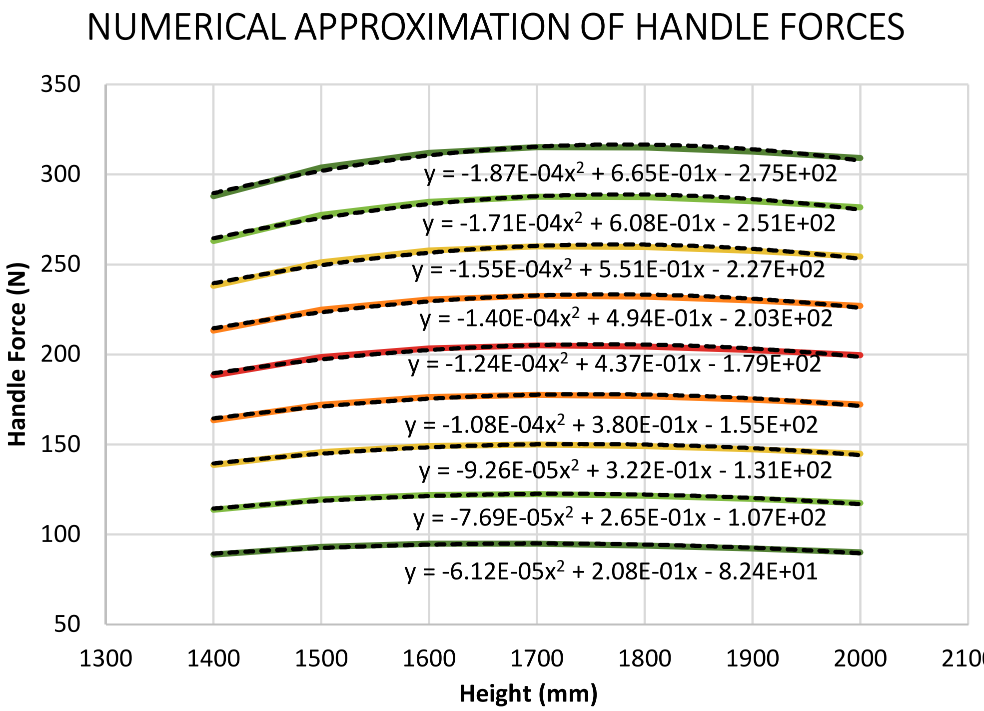


Figure 59 Developing the Polynomial Relationship between Height and Handle Force for a Variety of Weights

The values of the coefficients can then be linearly approximated using Equations 41, 42, and 43 to calculate handle force in Equation 44.

$$x_1 = -1.57 \times 10^{-6} W + 1.67 \times 10^{-6} \quad [41]$$

$$x_2 = 5.7 \times 10^{-3} W + 1.98 \times 10^{-2} \quad [42]$$

$$x_3 = -2.4099. W + 13.997 \quad [43]$$

$$x_1 . H^2 + x_2 . H + x_3 = HF \quad [44]$$

9.4 Testing and Results

To validate that the simulated trajectories, motion capture was used to assess the HTS3 mechanism. The motion capture and simulated trajectories were compared as shown in Figure 60, and found to be consistent. Unlike the results for HTS2, the active chest pad used for HTS3 testing does not allow the torso to bend; the hip is seen to track the simulated path very closely.

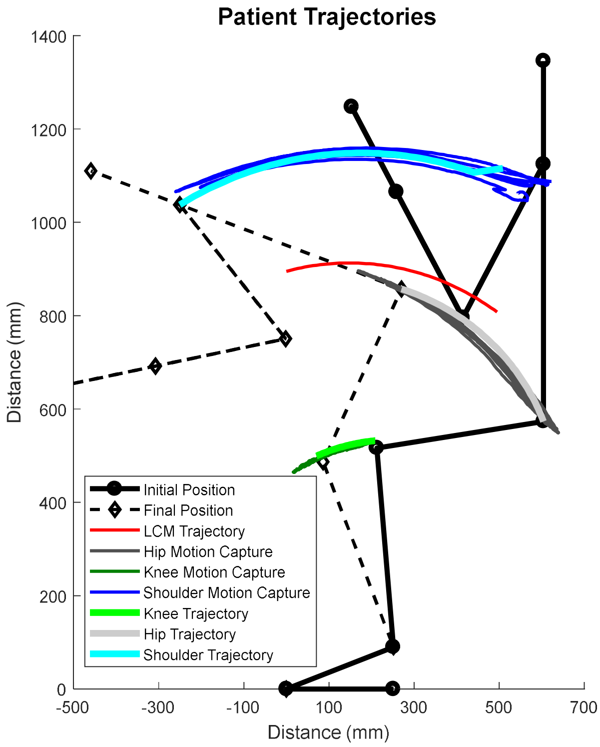


Figure 60 HTS3 Trajectory Comparison and Validation for a 1800 Millimetre, 87 Kilogram Patient

The height range of patients is limited to between 1400 and 1950 millimetres. To validate the handle force calculations three patients of known height and weight, as shown in Table 27, were lifted.

Table 27 Patient Characteristics for HTS3 Validation

Patient	Height (mm)	Weight (kg)
P1	1580	57
P2	1800	82
P3	1780	65

The secondary pivot arm length was noted, the standard active chest pad was fitted, and quasi-static handle testing was completed, with the results shown in Figure 61.

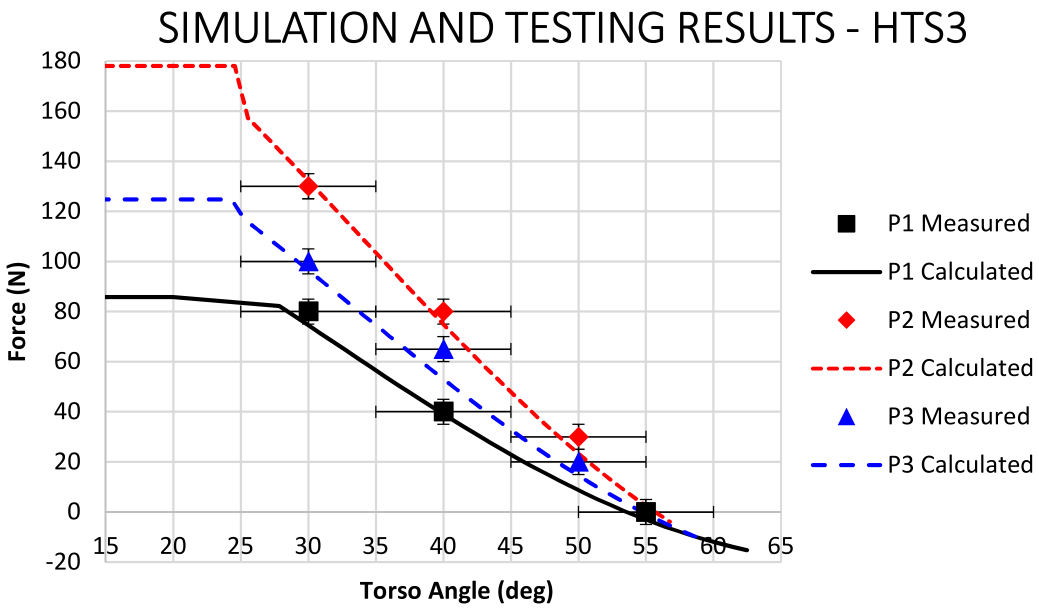
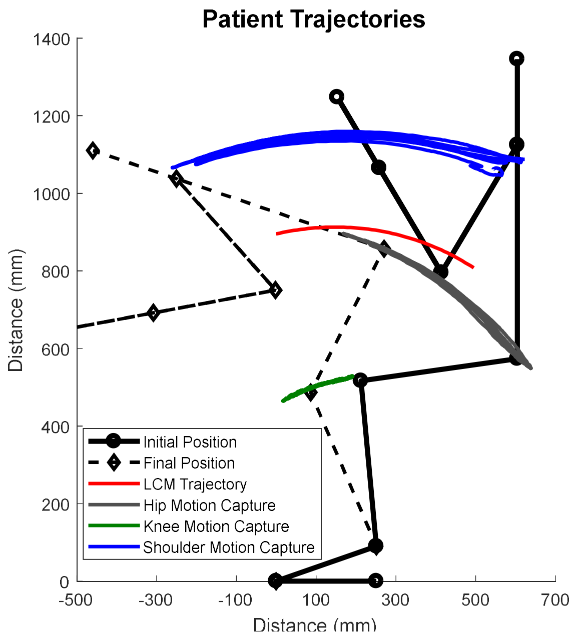


Figure 61 Simulation and Testing Result Comparison for HTS3 Mechanism

It is seen that the calculated and measured results closely match, with all measured data points within range of the calculated forces. Uncertainties from testing are shown using error bars of ± 5 Newtons and ± 5 degrees. The initial torso angle measurement was completed at 30 degrees to eliminate the possibility that the patient's weight had not been completely removed from the chair.

A study of the trajectories and handle forces of twelve patients of varied height and weight was completed using dynamic handle testing. A summary of the patient's results in shown in Figure 62, with comprehensive results shown in Appendix I.



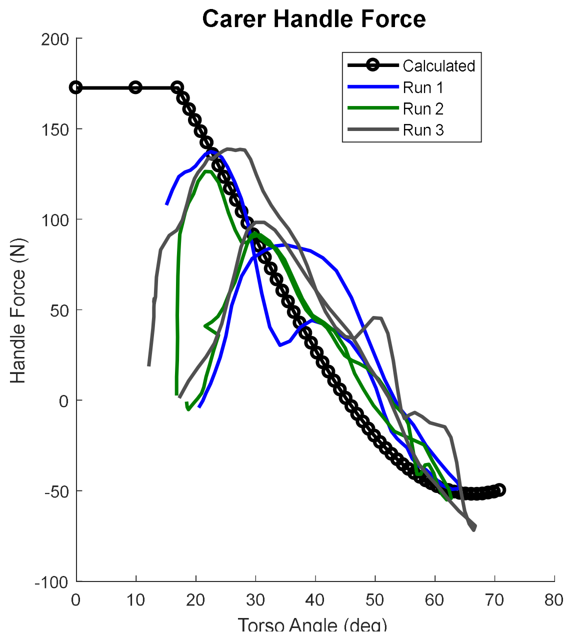


Figure 62 Dynamic Handle Force and Motion Capture Results for a 1800mm 87kg Patient

It is seen that the measured forces closely follow the predicted forces for all cases where the torso angle is larger than approximately 30 degrees. Prior to this, the forces are lower than expected. It is anticipated that this is due to the gradual application of the patient's weight to the mechanism as the chair or lifting surface will provide support to the patient until the patient's thigh fully loses contact with it. This is contrary to the simulation that relies on the assumption that the total patient weight is applied to the mechanism at the point the hip joint is raised from the chair.

It is noted that the carer handle forces never exceed the simulated handle force. For the case where the patient's seat is lower than the shank length, a 460 millimetre stool, it is seen that the forces were between 95 and 100 percent of the simulated force. This is contrary to transfers from higher surfaces, where the peak force value was seen to alter dramatically. The range in measured to simulated peak force varied from 45 to 100 percent. It is anticipated that this is due to the variation in the angle at which the patient's weight is fully loaded onto the mechanism. For a lower seat, the patient is raised when the torso angle is very acute. Conversely, for a high seat, the patient's weight is fully loaded onto the mechanism at a much larger angle. For most cases, the torso angle for the point at which the patient weight was totally supported by the mechanism when lifted from a standard 490 millimetre seat, was found to be between 16 and 22 degrees. For a 560 millimetre seat, the fully supported torso angle was found to range between 27 and 33 degrees.

Patients were asked not to aid the lift in any way during testing, to ensure fully passive lifts were completed. This allowed for the comparison of live and passive lifts to be completed. Figure 63 shows the impact a live lift has on decreasing the carer handle forces.

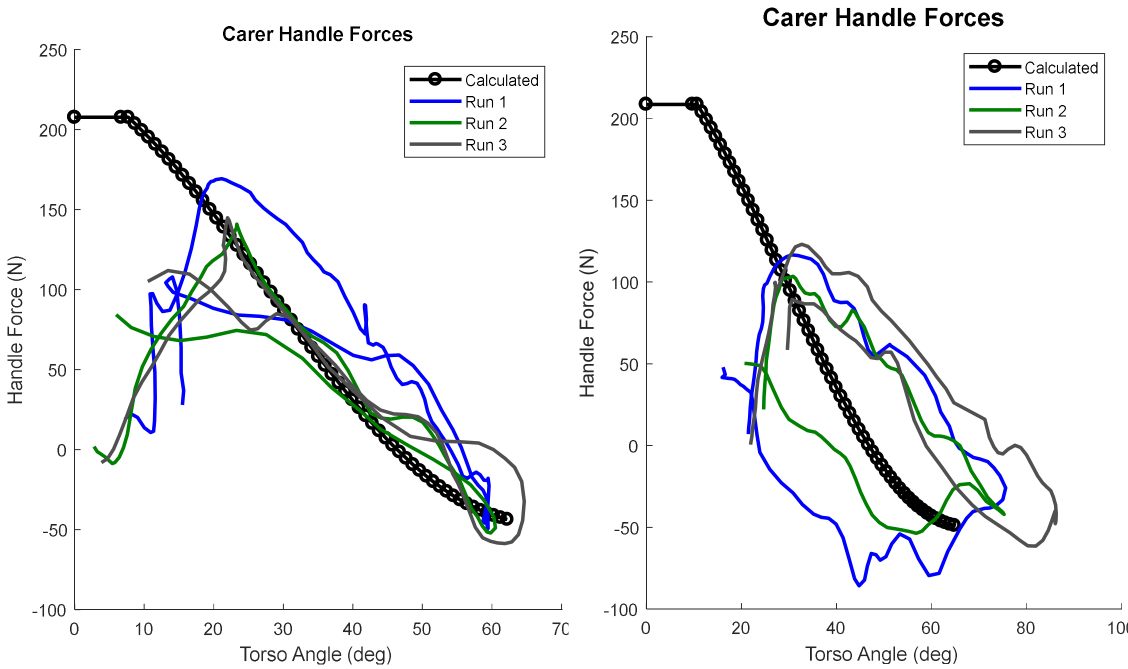


Figure 63 Comparison of Live (Left) and Passive (Right) Lifts

From this, it can be noted that handle forces are greatly reduced in the case of a live lift. The point at which the patient's weight is supported by the mechanism and how much input the patient applies during the lift are both factors that can not be effectively calculated or estimated. As such, it is anticipated that it is unlikely to be able to more accurately assess the peak force of a lift. The simulation and numerical approximation are useful as a guideline of an expected maximum force.

9.5 Summary

HTS3 addresses the main design issues of the Single Pivot, Tilting Chest Pad, and Little Blue mechanisms. The performance of the HTS3 mechanism DSR is 51 and is shown in Table 28 .

Table 28 Design Specifications, HTS3 Lifter Mechanism Performance Shown in Grey

Value \ Success Criteria	**Ease of Use**	**Carer Input**	**Safety**	**Stability**	**Cost**	**Manoeuvrability**	**Cognitive Requirement**
	Time taken for transfer (TT)	Carer force as percentage of body weight (CF)	Number of the following criteria met (MC)	Percentage of weight patient-supported (WB)	Estimated cost of device in NZD (DC)	Turning circle radius (TC)	Patient cognitive requirement (CR)
1	$10\,min < TT$	$80\% < CF$	• No carer forward back bending • **Load max 16kg** • **Locking mechanism** • **No twisting of carer torso** • **No dragging of patient** • **Force spread over large portion of patient's body** • **Varied surface height** • **No trap points** • **Transportable** • **Operation errors easily reversed**	$100\% \leq WB$	$\$5000 \leq DC$	$2.2m \leq TC$	$100\%\ CR$
2	$8\,min < TT \leq 10\,min$	$70\% < CF \leq 80\%$		$80\% < WB < 100\%$	$\$4000 < DC < \$5000$	$2.1m < TC < 2.2m$	
3	$6\,min < TT \leq 8\,min$	$60\% < CF \leq 70\%$		$70\% < WB \leq 80\%$	$\$3000 < DC \leq \$4000$	$2.0m < TC \leq 2.1m$	Able to stay alert and focused for entire lift
4	$5\,min < TT \leq 6\,min$	$50\% < CF \leq 60\%$		$60\% < WB \leq 70\%$	$\mathbf{\$2000 < DC \leq \$3000}$	$1.8m < TC \leq 2.0m$	
5	$4\,min < TT \leq 5\,min$	$40\% < CF \leq 50\%$		$50\% < WB \leq 60\%$	$\$1000 < DC \leq \$2000$	$1.6m < TC \leq 1.8m$	Able to follow a series of simple instructions
6	$\mathbf{3\,min < TT \leq 4\,min}$	$30\% < CF \leq 40\%$		$\mathbf{30\% < WB \leq 50\%}$	$\$750 < DC \leq \$1000$	$1.4m < TC \leq 1.6m$	
7	$2\,min < TT \leq 3\,min$	$20\% < CF \leq 30\%$		$20\% < WB \leq 30\%$	$\$500 < DC \leq \$750$	$1.2m < TC \leq 1.4m$	Able to follow a simple instruction
8	$1\,min < TT \leq 2\,min$	$\mathbf{10\% < CF \leq 20\%}$		$10\% < WB \leq 20\%$	$\$250 < DC \leq \$500$	$\mathbf{1.0m < TC \leq 1.2m}$	
9	$30\,s < TT \leq 1\,min$	$5\% < CF \leq 10\%$		$0\% < WB \leq 10\%$	$\$100 < DC \leq \$250$	$0.5m < TC \leq 1.0\,m$	$\mathbf{0\%\ CR}$
10	$TT \leq 30\,s$	$CF \leq 5\%$		$WB \leq 0\%$	$DC \leq \$100$	$TC \leq 0.5m$	

When compared with existing options within the lifting aid market, it is seen that the HTS3 concept meets all criteria to a satisfactory level. It should be noted that no other device assessed achieved such high scores in all sections of the evaluation matrix. As discussed previously in Section 2.6.6, transfer boards and mobile hoists bookend the assistive device market leaving a gap between the two extremes. It can be seen that the HTS3 fills this gap effectively. This result indicates the suitability of the HTS3 for a large proportion of the market. From this, it can be anticipated that the HTS3 would be beneficial in not only commercial care situations, but also domestic situations, supporting the desire of the elderly to stay in their own homes.

10 Miscellaneous Concepts

10.1 Four Bar Linkage

In trying to optimise and develop a suitable lifter to enter the healthcare market, many prototypes have been created. Previously created lifters include the Four Bar Linkage Lifter shown in Figure 64.

Figure 64 Four Bar Linkage Lifter

The four bar linkage system was developed to mimic the Little Blue Lifter, but with a less invasive mechanism. The concept was developed by considering how the mechanism could provide a similar lifting path to the Little Blue Lifter but using a mechanism located in front of the patient's knees, nearer the carer. The geometry of the Four Bar Linkage Mechanism is shown in Figure 65.

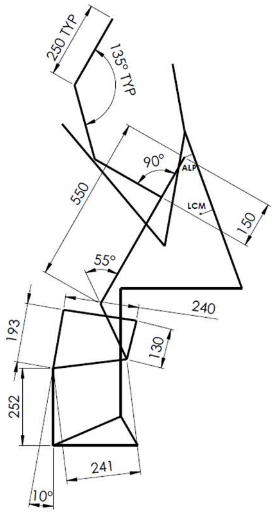


Figure 65 Geometry and Dimensions of Four Bar Linkage Mechanism

The mechanism was found to be unstable in the lifted position, with the carer needing to apply force to ensure the patient remained lifted. This occurred consistently, even when the entire lifter was rotated on the chassis as shown in Figure 64. The performance of the Four Bar Linkage DSR is 43 and is shown in Table 29. It was found that the mechanism did not allow for the desired trajectories. It was decided to focus on the Slider Plate discussed in Section 10.2 to allow for more freedom in trajectory choice.

Table 29 Design Specifications, Four Bar Linkage Mechanism Performance Shown in Grey

Success Criteria / Value	Ease of Use	Carer Input	Safety	Stability	Cost	Manoeuvrability	Cognitive Requirement
	Time taken for transfer (TT)	Carer force as percentage of body weight (CF)	Number of the following criteria met (MC)	Percentage of weight patient-supported (WB)	Estimated cost of device in NZD (DC)	Turning circle radius (TC)	Patient cognitive requirement (CR)
1	$10\ min < TT$	$80\% < CF$	• No carer forward back bending • **Load max 16kg** • Locking mechanism • **No twisting of carer torso** • No dragging of patient • Force spread over large portion of patient's body • **Varied surface height** • No trap points • **Transportable** • Operation errors easily reversed	$100\% \leq WB$	$\$5000 \leq DC$	$2.2m \leq TC$	$100\%\ CR$
2	$8\ min < TT \leq 10\ min$	$70\% < CF \leq 80\%$		$80\% < WB < 100\%$	$\$4000 < DC < \$5000$	$2.1m < TC < 2.2m$	
3	$6\ min < TT \leq 8\ min$	$60\% < CF \leq 70\%$		$70\% < WB \leq 80\%$	$\$3000 < DC \leq \$4000$	$2.0m < TC \leq 2.1m$	Able to stay alert and focused for entire lift
4	$5\ min < TT \leq 6\ min$	$50\% < CF \leq 60\%$		$60\% < WB \leq 70\%$	$\mathbf{\$2000 < DC \leq \$3000}$	$1.8m < TC \leq 2.0m$	
5	$4\ min < TT \leq 5\ min$	$40\% < CF \leq 50\%$		$50\% < WB \leq 60\%$	$\$1000 < DC \leq \$2000$	$1.6m < TC \leq 1.8m$	**Able to follow a series of simple instructions**
6	$3\ min < TT \leq 4\ min$	$30\% < CF \leq 40\%$		$\mathbf{30\% < WB \leq 50\%}$	$\$750 < DC \leq \$1000$	$1.4m < TC \leq 1.6m$	
7	$2\ min < TT \leq 3\ min$	$20\% < CF \leq 30\%$		$20\% < WB \leq 30\%$	$\$500 < DC \leq \$750$	$1.2m < TC \leq 1.4m$	Able to follow a simple instruction
8	$\mathbf{1\ min < TT \leq 2\ min}$	$\mathbf{10\% < CF \leq 20\%}$		$10\% < WB \leq 20\%$	$\$250 < DC \leq \$500$	$\mathbf{1.0m < TC \leq 1.2m}$	
9	$30\ s < TT \leq 1\ min$	$5\% < CF \leq 10\%$		$0\% < WB \leq 10\%$	$\$100 < DC \leq \$250$	$0.5m < TC \leq 1.0\ m$	$0\%\ CR$
10	$TT \leq 30\ s$	$CF \leq 5\%$		$WB \leq 0\%$	$DC \leq \$100$	$TC \leq 0.5m$	

10.2 Slider Plate

The Slider Plate mechanism was developed to assess the feasibility of a zero or minimal force lift. The mechanism and chosen trajectory are shown in Figure 66.

Figure 66 Slider Plate Mechanism and Trajectory

It was considered that a slider plate option could be used to test trajectories incompatible with the four bar linkage system. From testing it was found that as the linkage lengths increased, the handle force decreased. The trajectory shown above was developed to decrease the initial handle forces, provide a low force, consistent lift, and hold the patient in a stable transport position. The geometry of the Single Slider Plate Mechanism is shown in Figure 67.

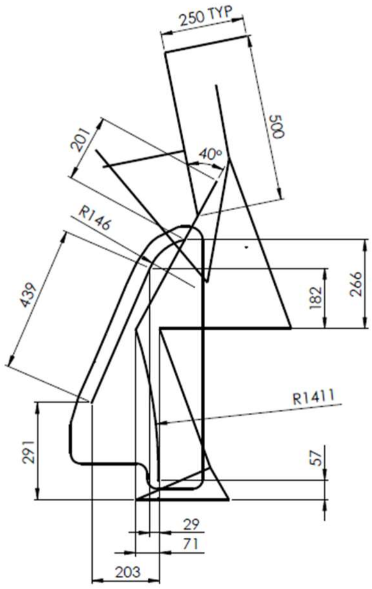


Figure 67 Geometry and Dimensions of Slider Plate Mechanism

While lower than previously measured, the forces were still found to exceed 60 Newtons. There were also obvious inherent flaws with this design, most notably the safety of a slider plate arrangement. It was also found that the moments created by the two slots increased forces. While it was found that the forces were low, these forces were not as reduced as expected. This is largely due to the friction involved in the double slot motion and the moment caused by any uneven loading of the patient. The mechanism was also bulky and intimidating to patients.

This design was developed to utilise the findings of the Four Bar Linkage mechanism to lower lifter forces and increase ease of use. The performance of the Slider Plate Linkage DSR is 38 and is shown in Table 30 .

Table 30 Design Specifications, Slider Plate Mechanism Performance Shown in Grey

Value \ Success Criteria	Ease of Use	Carer Input	Safety	Stability	Cost	Manoeuvrability	Cognitive Requirement
	Time taken for transfer (TT)	Carer force as percentage of body weight (CF)	Number of the following criteria met (MC)	Percentage of weight patient-supported (WB)	Estimated cost of device in NZD (DC)	Turning circle radius (TC)	Patient cognitive requirement (CR)
1	$10\ min < TT$	$80\% < CF$	• No carer forward back bending • **Load max 16kg** • Locking mechanism • No twisting of carer torso • No dragging of patient • Force spread over large portion of patient's body • Varied surface height • No trap points • Transportable • Operation errors easily reversed	$100\% \leq WB$	$\$5000 \leq DC$	$2.2m \leq TC$	$100\%\ CR$
2	$8\ min < TT \leq 10\ min$	$70\% < CF \leq 80\%$		$80\% < WB < 100\%$	$\$4000 < DC < \$5000$	$2.1m < TC < 2.2m$	
3	$6\ min < TT \leq 8\ min$	$60\% < CF \leq 70\%$		$70\% < WB \leq 80\%$	$\$3000 < DC \leq \$4000$	$2.0m < TC \leq 2.1m$	**Able to stay alert and focused for entire lift**
4	$5\ min < TT \leq 6\ min$	$50\% < CF \leq 60\%$		$60\% < WB \leq 70\%$	$\boldsymbol{\$2000 < DC \leq \$3000}$	$1.8m < TC \leq 2.0m$	
5	$4\ min < TT \leq 5\ min$	$40\% < CF \leq 50\%$		$50\% < WB \leq 60\%$	$\$1000 < DC \leq \$2000$	$1.6m < TC \leq 1.8m$	Able to follow a series of simple instructions
6	$3\ min < TT \leq 4\ min$	$30\% < CF \leq 40\%$		$\boldsymbol{30\% < WB \leq 50\%}$	$\$750 < DC \leq \$1000$	$1.4m < TC \leq 1.6m$	
7	$2\ min < TT \leq 3\ min$	$20\% < CF \leq 30\%$		$20\% < WB \leq 30\%$	$\$500 < DC \leq \$750$	$\boldsymbol{1.2m < TC \leq 1.4m}$	Able to follow a simple instruction
8	$\boldsymbol{1\ min < TT \leq 2\ min}$	$10\% < CF \leq 20\%$		$10\% < WB \leq 20\%$	$\$250 < DC \leq \$500$	$1.0m < TC \leq 1.2m$	
9	$30\ s < TT \leq 1\ min$	$\boldsymbol{5\% < CF \leq 10\%}$		$0\% < WB \leq 10\%$	$\$100 < DC \leq \$250$	$0.5m < TC \leq 1.0\ m$	$0\%\ CR$
10	$TT \leq 30\ s$	$CF \leq 5\%$		$WB \leq 0\%$	$DC \leq \$100$	$TC \leq 0.5m$	

10.3 Sliding Pivot

The Sliding Pivot mechanism was developed from the HTS2 concept. It was developed to assess the feasibility of removing the initial high handle force of a lift and the impact of this on the change in height of the LCM. The mechanism is shown in Figure 68.

Figure 68 Sliding Pivot Mechanism

It was considered that, if the initial high forces in a lift could be removed, the overall forces during a lift could be increased. The key benefit of this would be the ability to raise the LCM higher than previously, without increasing the peak force. The geometry of the Sliding Pivot Mechanism is shown in Figure 69.

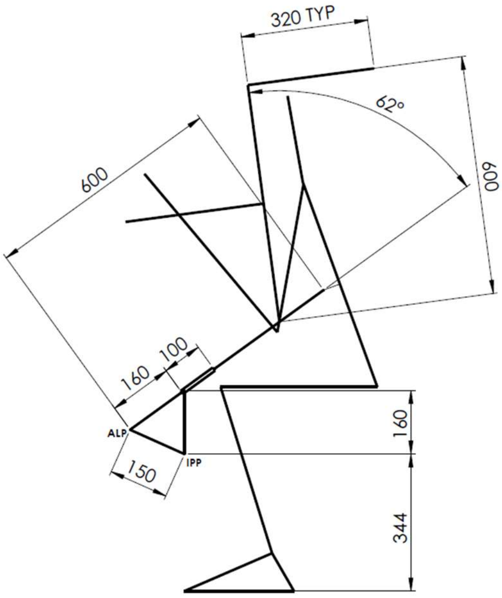


Figure 69 Geometry and Dimensions of Sliding Pivot Mechanism

Handle forces were found to be lower when correctly set up for a patient. However, small variations in the patient resulted in large force differences. It was found that the sensitivity of the mechanism to small variations of patient's LCM, leg angles, torso angles, and seat heights was too large to allow the mechanism to be a successful solution. The low forces, increased LCM height, and increased comfort levels do indicate that with more consideration, this mechanism could be greatly beneficial and would benefit from further development. It is also possible that, due to the suitability over a small range of patients, this mechanism could be developed as bespoke residential lifting aids.

This design was developed to utilise the findings of the Tilting Chest Pad mechanism to lower lifter forces and increase LCM height. The performance of the Sliding Pivot Linkage DSR is 40 and is shown in Table 31 .

Table 31 Design Specifications, Sliding Pivot Mechanism Performance Shown in Grey

Success Criteria / Value	Ease of Use	Carer Input	Safety	Stability	Cost	Manoeuvrability	Cognitive Requirement
	Time taken for transfer (TT)	Carer force as percentage of body weight (CF)	Number of the following criteria met (MC)	Percentage of weight patient-supported (WB)	Estimated cost of device in NZD (DC)	Turning circle radius (TC)	Patient cognitive requirement (CR)
1	$10\ min < TT$	$80\% < CF$	• No carer forward back bending • Load max 16kg • Locking mechanism • **No twisting of carer torso** • No dragging of patient • **Force spread over large portion of patient's body** • **Varied surface height** • No trap points • Transportable • **Operation errors easily reversed**	$100\% \leq WB$	$\$5000 \leq DC$	$2.2m \leq TC$	$100\%\ CR$
2	$8\ min < TT \leq 10\ min$	$70\% < CF \leq 80\%$		$80\% < WB < 100\%$	$\$4000 < DC < \$5000$	$2.1m < TC < 2.2m$	
3	$6\ min < TT \leq 8\ min$	$60\% < CF \leq 70\%$		$70\% < WB \leq 80\%$	$\$3000 < DC \leq \$4000$	$2.0m < TC \leq 2.1m$	Able to stay alert and focused for entire lift
4	$5\ min < TT \leq 6\ min$	$50\% < CF \leq 60\%$		$60\% < WB \leq 70\%$	$\$2000 < DC \leq \$3000$	$1.8m < TC \leq 2.0m$	
5	$4\ min < TT \leq 5\ min$	$40\% < CF \leq 50\%$		$50\% < WB \leq 60\%$	$\mathbf{\$1000 < DC \leq \$2000}$	$1.6m < TC \leq 1.8m$	**Able to follow a series of simple instructions**
6	$3\ min < TT \leq 4\ min$	$30\% < CF \leq 40\%$		$\mathbf{30\% < WB \leq 50\%}$	$\$750 < DC \leq \$1000$	$\mathbf{1.4m < TC \leq 1.6m}$	
7	$\mathbf{2\ min < TT \leq 3\ min}$	$\mathbf{20\% < CF \leq 30\%}$		$20\% < WB \leq 30\%$	$\$500 < DC \leq \$750$	$1.2m < TC \leq 1.4m$	Able to follow a simple instruction
8	$1\ min < TT \leq 2\ min$	$10\% < CF \leq 20\%$		$10\% < WB \leq 20\%$	$\$250 < DC \leq \$500$	$1.0m < TC \leq 1.2m$	
9	$30\ s < TT \leq 1\ min$	$5\% < CF \leq 10\%$		$0\% < WB \leq 10\%$	$\$100 < DC \leq \$250$	$0.5m < TC \leq 1.0\ m$	$0\%\ CR$
10	$TT \leq 30\ s$	$CF \leq 5\%$		$WB \leq 0\%$	$DC \leq \$100$	$TC \leq 0.5m$	

11 Discussion

11.1 Overview

The theory of forces involved in patient handling was assessed to evaluate and develop a fully mechanical patient lifter. This theory was extended to include the forces present in lifting mechanisms to allow for prediction and improvement of mechanism handle forces. Validation of the theory was completed through two stages of testing. Preliminary testing was completed using a spring balance, with more extensive and accurate secondary testing being completed using a load cell and motion capture software. The implication of patient variation on handle forces was also modelled and tested. The findings from this were used to develop the lifting mechanism to ensure the performance was consistent over a range of patients. The success of the lifting mechanism was assessed through the use of a developed evaluation matrix.

11.2 New Zealand Health Survey Data

As discussed previously, patient height, weight, and centre of mass position have a large impact on the forces and patient trajectories, while more subtle patient variations such as thigh and torso length also figure. The New Zealand Health Survey was used to provide a more accurate representation of the New Zealand population, and specifically the aged 75 and over population's anthropometric data. The height and weight of both males and females decreases with age. Conversely, it is seen that the waist circumference increases with age.

The New Zealand Health Survey was also used to assess a possible market for the lifter through the assessment of responses to self-reported health questions. By focussing on these questions, it was possible to eliminate respondents unlikely to require mobility aids. Assessment showed that 93.5 percent of the population were deemed unlikely to require a mobility aid, with the most common exclusions being acceptable quality of self-reported health, high levels of exercise, and living alone.

These findings indicate that 6.5 percent, or 258,000 people may benefit from using a patient lifter. The findings from the NZ Health Survey data were also used to assess possible patient characteristics and restrictions. The restrictions for the HTS3 mechanism are summarised in Table 32.

Table 32 Percentage of Relevant Populations within HTS3 Patient Restrictions

Measurement	Patient Restriction	Age Group	Gender	Percent Within Range (%)
Weight (kg)	120	All Ages	Male	94.7
			Female	96.7
		Aged 75 and Over	Male	98.9
			Female	99.5
Height (mm)	2000	All Ages	Male	99.7
			Female	100
		Aged 75 and Over	Male	100
			Female	100
Waist Circumference (mm)	1200	All Ages	Male	92.3
			Female	94.9
		Aged 75 and Over	Male	91.5
			Female	95.6

11.3 ANTHROPOMETRIC VARIATIONS

The key to understanding the implications of patient variability on handle forces is in understanding the forces within the patient and how these are transferred to the mechanism. Development of the patient forces used a quasi-static approach where momentum is ignored and the lift is assumed to be relatively slow. This is summarised in Appendix J.

As discussed previously, it was found that the position of the LCM is critical in understanding the forces applied to and by the patient. Therefore, anything that influences of the LCM position will noticeably effect the handle force. This is especially true for any adjustment to the LCM position in the horizontal direction. The key influences on handle force, and the causes for this, are shown in Table 33.

Table 33 Implications of Patient Variations on Handle Forces

	Patient	Mechanism	Forces
Height Increased	Thigh length increases causing torso, and LCM, further from knees	Secondary pivot arm length increased to ensure chest pad contact maintained	Increase as steeper initial lift due to increased secondary pivot arm length
Weight Increased	Larger weight force applied at LCM	-	Increase proportional weight increase
Waist Circumference Increased	LCM moves closer to contact point of patient with chest pad	Secondary pivot arm decreased to allow for larger waist	Decrease as chest forces lower due to LCM shift and trajectory gradient decrease
LCM Lowered	LCM moves further from contact point of patient with chest pad	-	Increase proportional the LCM adjustment
Shank Length Increased	Hip angle more acute, LCM trajectory steeper	-	Increase dependant on shank length to chair height ratio
Thigh Length Increased	LCM moves further from contact point of patient with chest pad	Secondary pivot arm length is increased to ensure chest pad contact maintained	Increase as steeper initial lift due to increased secondary pivot arm length
Torso Length Increased	LCM trajectory gradient decreased at end of lift	-	Initial force constant, with decreased forces at transport position
Thigh Weight Increased	LCM lowered, moving further forward	-	Chest forces lower due to altered LCM position and gradient
Torso Weight Increased	LCM raised and moved closer to spine	-	Increase respective of chest forces due to LCM shift and trajectory gradient increase

11.4 Mechanism Development

As stated previously, the difference between the angle of the ALP trajectory, ε, and the resultant chest pad force angle, δ, greatly influences the handle force the carer is required to apply. This is key when combining patient and mechanism theories to allow for development of low force lifting. The difference between δ and ε and the magnitude of FC are combined to find FEI. Obviously, the magnitude of the handle force is also influenced by the length of the handle.

Initially, the focus of this project was to generate a mechanism that would allow a patient lift with effectively no carer input force. However, after the HTS2 was manufactured, it was found that forcing the LCM along a trajectory of low gradient was uncomfortable for the patient. The discomfort originated from the noticeable decrease in the thigh to torso angle. It was found that this level of decrease was especially uncomfortable in elderly patients, as joint flexibility decreases with age. In some cases, patients were not able to complete the lift due to the required torso to thigh angle.

It was also noticed the shank to thigh angle greatly increased during the lift. This was also of concern as it was found to cause hamstring strain within patients. It was identified many patients lifted their heels off the footplate to counteract this. As such, it was decided the focus should be adjusted to include patient comfort. The balance in requirements when developing mechanisms is shown in Figure 70 .

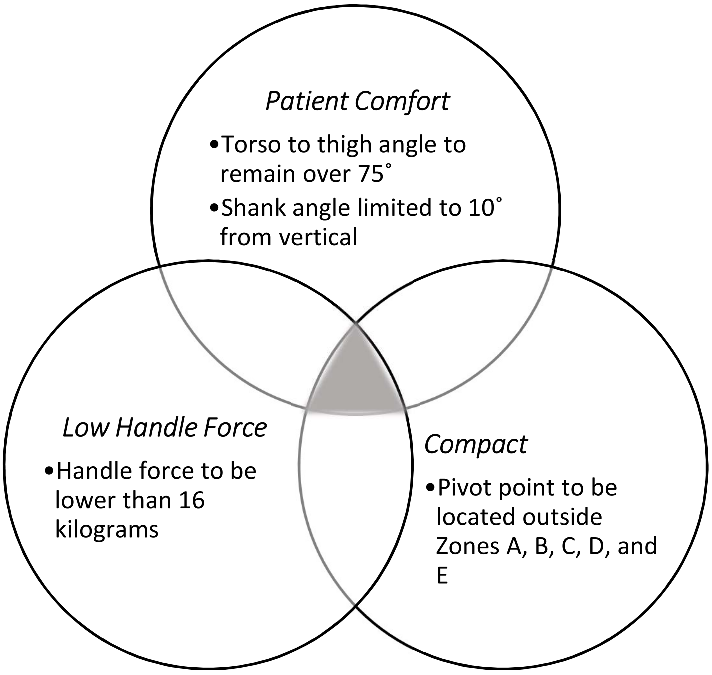


Figure 70 Venn Diagram Outlining the Focus for Balance in Mechanism Design

Figure 70 illustrates that there is room to achieve patient comfort by ensuring the torso to thigh angle remains over 75 degrees while the shank angle is kept below 10 degrees. The shaded area of this diagram indicates a suitable balance between the desired lifter features. It is also known, from theory, that increasing the steepness of the LCM trajectory increases the handle forces, as the difference between δ and ε becomes larger. Therefore, the key to developing a low-force, comfortable lifter is to ensure a gradual raising of the LCM.

As opposed to an initial rapid rise, the initial forces will continue to be low while the LCM will be raised enough to allow patient comfort. From an assessment of patient and LCM trajectories, HTS3 was developed with the objective of achieving patient comfort and limiting handle forces.

As discussed previously in Section 9.1, the concept for the HTS3 is based on the HTS2 tilting chest pad mechanism with adjusted linkage lengths. The HTS3 allows for a range of patient variations through an adjustable secondary pivot arm length, R2. A chest strap was also added to ensure security and stability of the patient throughout the lift.

The HTS3 provides a suitable low force lift paired with high patient comfort due to the trajectory. The mechanism is simple and easily maintained. The selected pivot points provide good clearance of furniture and allows the patient to retrieved from, and replaced, far back in chairs. This results in no repositioning of the patient once a transfer has taken place. This is highly beneficial and unique within the market. The mechanism can also be used for rehabilitation as patients can progressively aid in the lift, redeveloping their weight bearing capabilities.

A key limitation of the HTS3 is the rolling resistance. While mechanical advantage can be used to decrease forces when lifting patients, the resistance on compliant and carpeted surfaces is still comparative to a mobile hoist. While it is anticipated this is not an issue, it should be noted that studies on push and manoeuvring forces for mobile and standing hoists have been found to have force spikes over 16 kilograms, or 157 Newtons.

It is recommended that development of the handle be completed. While the increased length of the handle is beneficial in reducing forces, the increased length also increases the footprint of the lifter mechanism. It is possible that a telescoping or folding handle would be beneficial to manoeuvring in smaller areas.

A more in-depth assessment of the forces within the lifter may allow for some of the members to be decreased in size. This would aid in decreasing materials cost.

It should be noted that 25 percent of the total population were removed when considering the possible lifter market due to living alone. A large percent of these single person dwellings were elderly. This highlights the possibility for marketing a device to aid patients in sit to stand lifts without a carer. This could potentially increase the ability for elderly to “age in place”.

Another key market area is bariatric patients. It is possible that a minimum force trajectory lift would allow for a bariatric patient, over 150 kilograms, to be manually lifted. This would provide a simple and lower cost method of bariatric patient handling. The key limitation of this mechanism will be the increased rolling resistance due to increased patient weight.

While the HTS3 does not achieve the highest scores in every specification category, it can be seen that it appears to be the most well rounded of the concepts. It is anticipated that this indicates the device would be well suited to a larger range of patients and provide a useful intermediate step between transfer boards and mobile hoists.

The progression of the lifter development, and features and limitations can be seen in Figure 71.

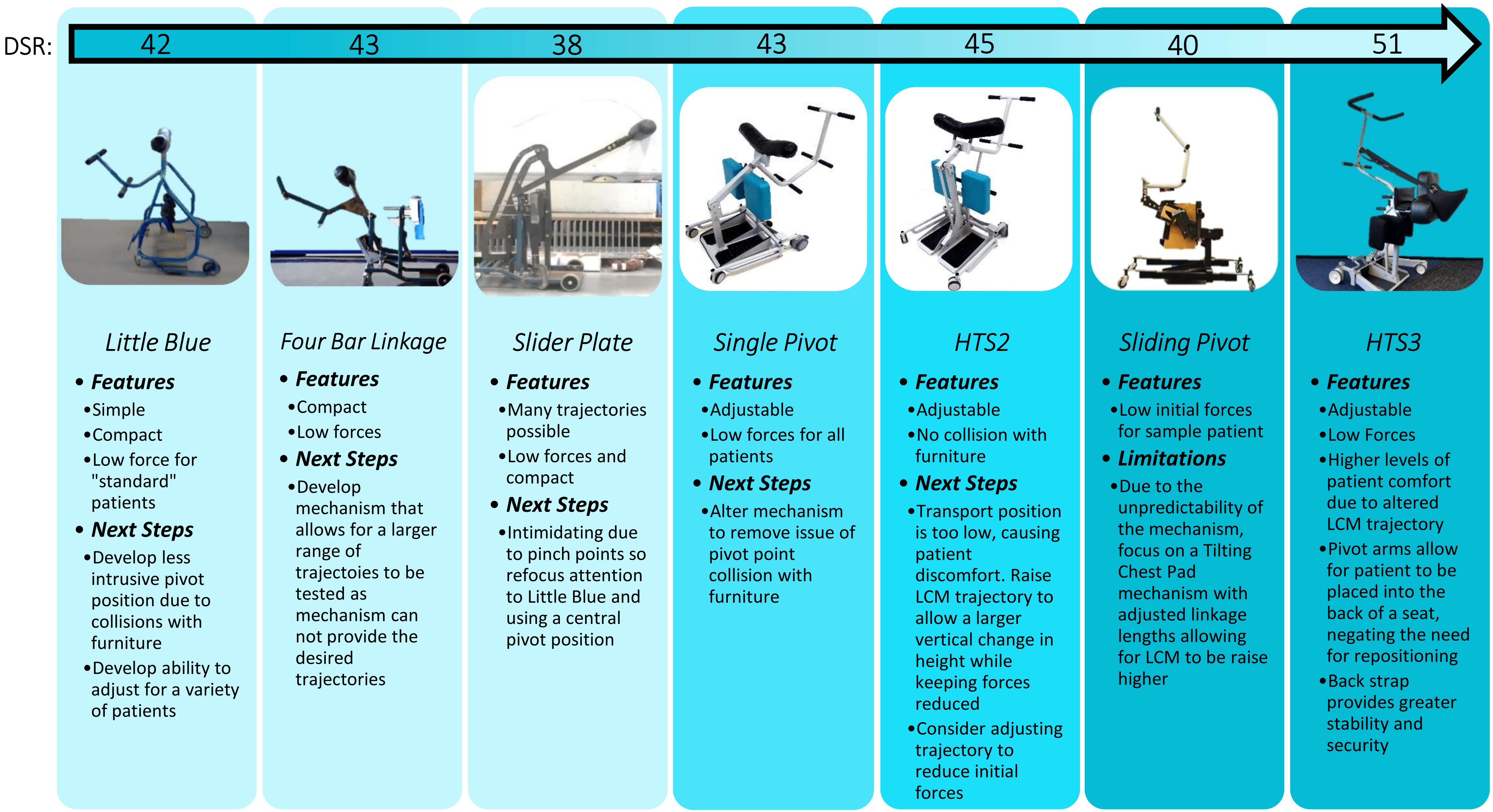


Figure 71 Progression of Mechanism Development

An assessment of the HTS3 compared to existing lifters in the market is shown in Table 34.

Table 34 Performance of HTS3 and Existing Lifting Solutions

	Mobile Hoists	Standing Hoists	Transfer Belts	Transfer Boards	Manual Lifters	HTS3
Ease of Use	5	5	8	8	7	6
Carer Input	8	7	2	10	6	8
Safety	8	7	3	5	5	9
Stability	10	3	2	3	3	6
Cost	2	2	10	9	3	4
Manoeuvrability	7	7	10	9	8	8
Cognitive Requirement	10	5	3	1	6	9
Total	**50**	**36**	**38**	**45**	**38**	**51**

It can be seen that HTS3 has a highest overall score of any existing or developed solution. It is noted that the lowest score the HTS3 has received is for cost, although this is reasonable when compared to mobile and standing hoists. It is anticipated that the main competition for the HTS3 will be standing hoists and manual lifters, both of which have much lower overall comparative scores as seen in Figure 72.

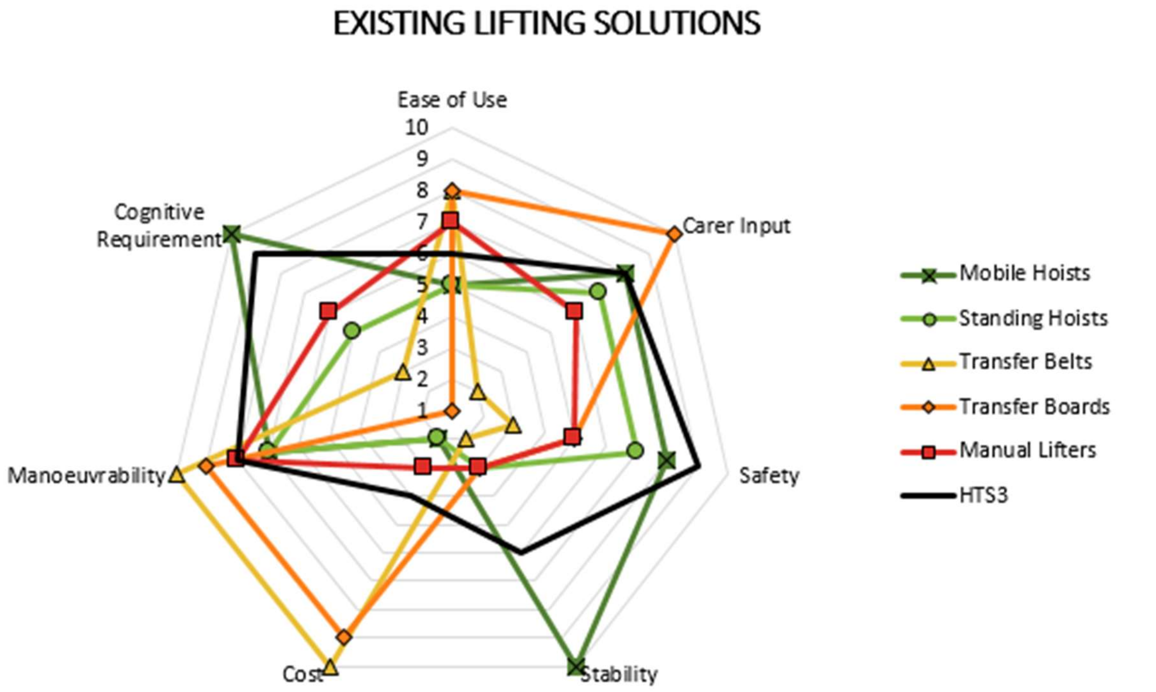


Figure 72 Comparison of HTS3 Performance with Existing Patient Lifting Devices

This clearly shows that HTS3 outperforms both manual lifters and standing hoists in most areas of evaluation. It is evident that the HTS3 provides an intermediate step between transfer boards and mobile hoists.

11.5 Patent Region

As discussed previously, the initial focus of this research was to assess the potential of developing a mechanism that greatly reduced carer handle forces. Utilising patient and lifter theory, it was found that to reduce forces without altering the mechanism geometry, the chest pad resultant force needed to be reduced.

The only feasible way to alter the chest pad resultant force for a specific patient is to reduce the magnitude of the component of the resultant chest pad force not acting along the ALP trajectory. This can be completed through decreasing the angle between the line normal to the ALP trajectory and the resultant chest pad force angle. From this, it was found that when this angle decreased to zero as discussed in Section 3.3, the carer input force was also reduced to zero.

Throughout this project, the focus has altered to provide a more comfortable lift for the patient, at the expense of low handle forces. With a company focus on potential markets within the patient handling sector, it was decided to assess a potential patentable trajectory region. A useful patentable region was defined as the region between the HTS3 shoulder, or jugular notch, trajectory, and the theoretical zero force shoulder or jugular notch trajectory. It is anticipated that this region is beneficial, as it includes the HTS3 path and the theoretical zero force path. The HTS3 path has been carefully developed and is a suitable balance of patient comfort and low handle forces. The theoretical zero force path is also advantageous as it allows a very large patient to be lifted with theoretically no carer input. These two trajectories have been defined as the maximum and minimum range of the region and are shown in Figure 73.

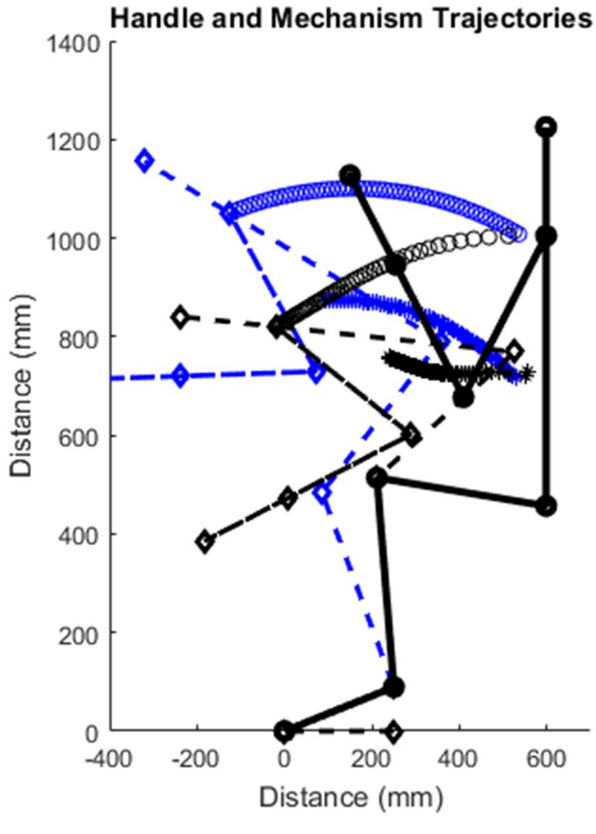


Figure 73 Patent Region between HTS3 Trajectory (Blue) and Zero Force Trajectory (Black)

The upper and lower limits of the regions can be calculated using Equations 45 and 46 respectively when the tip of the patient’s toe is located at (0,0).

$$y = -6.17 \times 10^{-4}x^2 + 0.316x + 1080 \quad [45]$$

$$y = -4.31 \times 10^{-4}x^2 + 0.496x + 903 \quad [46]$$

The initial value of x in these cases is defined as 36 percent of the total height of the patient in millimetres. This decreases to x equal to -200 to enclose the patentable region as shown in Figure 74.

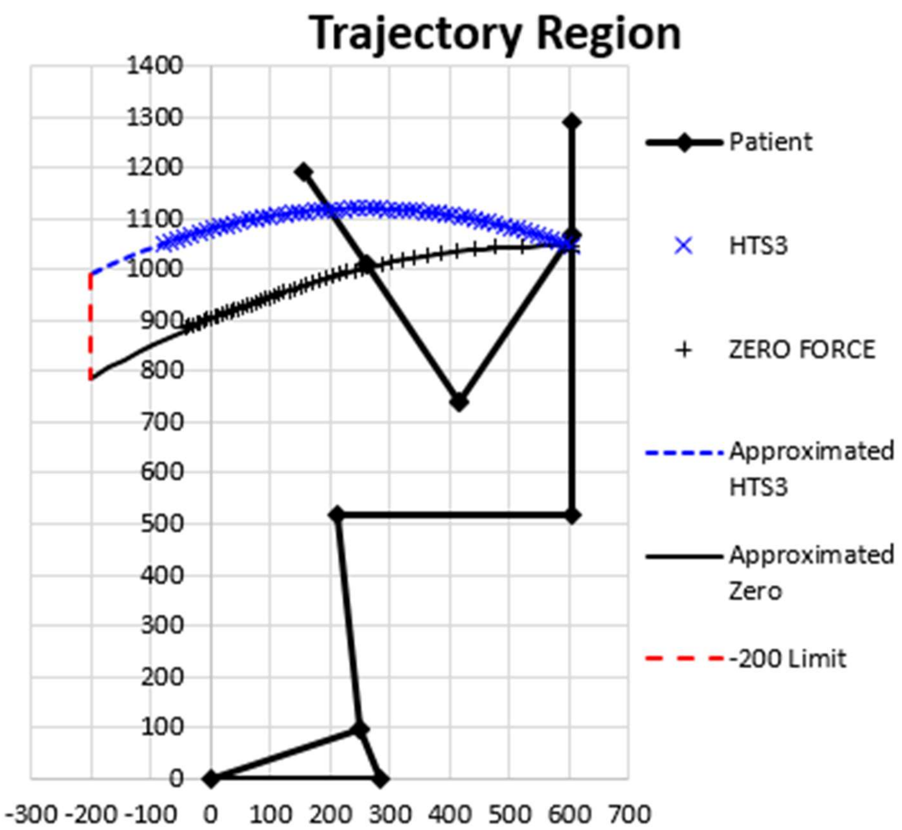


Figure 74 Patent Region Defined

Obviously, the location of this region will be dependent on the seat height as well as patient height. To adjust this region for seat height, the change in seat height from a standard 490 millimetre seat must be added to the trajectory. For example, a 550 millimetre seat would raise the entire region by 60 millimetres whereas a 450 millimetre seat would lower the entire region by 40 millimetres.

12 Conclusion

To address New Zealand's ageing population and the increase of demand on nursing staff and health services, a fully-mechanical patient lifter has been developed. It is anticipated that the simple, intuitive, and effective design of the lifter will allow it to become an intermediate step within the patient handling market, bridging the gap between transfer boards and mobile hoists.

A strong understanding of the forces involved during a patient lift, along with processed results from the New Zealand Health Survey, have aided in developing a low force lifter with the potential to greatly reduce the time and effort required to complete patient transfers. The developed mechanism utilises two pivot positions to achieve a change in Lifted Centre of Mass height of approximately 120 millimetres, with acceptably low forces.

13 REFERENCES

Abraham, B. B., & Johnson, G. R. (2010). A quasi-static state examination of handle forces and translational acceleration at impending planar motion for the four-caster manually manoeuvred vehicle. *Proceedings of the Institute of Mechanical Engineers, Part K: Journal of Multi-Body Dynamics, 224*, 143-156.

Accident Compensation Corporation. (2003). *The New Zealand Patient Handling Guidelines.* Wellington, New Zealand: Accident Compensation Corporation.

Accident Compensation Corporation. (2012). *Moving and Handling People Guidelines.* Wellington, New Zealand: Accident Compensation Corporation.

Alamgir, H., Li, O. W., Yu, S., Gorman, E., Fast, C., & Kidd, C. (2009). Evaluation of ceiling lifts: Transfer time, patient comfort and staff perceptions. *Injury*, 987-992.

Andersen, L. L., Fallentin, N., Thorsen, S. V., & Holtermann, A. (2016). Physical workload and risk of long-term sickness absence in the general working population and among blue-collar workers: prospective cohort study with register follow-up. *Occupational and Environmental Medicine*, 246-253.

Barbareschi, G., Cheng, T.-J., & Holloway, C. (2018). Effect of technique and transfer board use on the performance of wheelchair transfers. *Healthcare Technology Letters, 5*(2), 76-80.

Beard, J. R., & Bloom, D. E. (2015). Towards a comprehensive public health response to population ageing. *The Lancet*, 658-661.

Borner, H. E. (2008). *Evaluating Safe Patient Handling Systems: Is There a Better Way?* Wellington, New Zealand: Victoria University of Wellington.

Buckle, R. A., & Creedy, J. (2014). Population ageing and long-run fiscal sustainability in New Zealand. *New Zealand Economic Papers, 48*, 105-110.

Bureau of Labor Statistics. (2018). *Employment Projections and Occupational Outlook Handbook.* Washington DC; United States of America: U.S. Department of Labor.

Carey, D. (1999). *Coping with Population Ageing in Australia.* Paris, France: OECD Publishing.

Centers for Disease Control and Prevention. (2000). *Home Health - Data Highlights 2000.* Atlanta, United States of America: U.S. Department of Health and Human Services.

Collins, J. W., Nelson, A., & Sublet, V. (2006). *Safe Lifting and Movement of Nursing Home Residents.* Cincinnati; United States of America: National Institute for Occupational Safety and Health.

Cornwall, J., & Davey, J. A. (2004). *Impact of Population Ageing in New Zealand on the Demand for Health and Disability Support Services, and Workforce Implications.* Wellington, New Zealand: Ministry of Health.

de Leva, P. (1996). Adjustments to Zatsiorsky-Seluyanov's Segment Inertia Parameters. *Journal of Biomechanics*, 1223-1230.

Dutta, T., Holliday, P. J., Gorski, S. M., Baharvandy, M. S., & Fernie, G. R. (2012). A biomechanical assessment of floor and overhead lifts using one or two caregivers for patient transfers. *Applied Ergonomics*, 521-531.

Enos, L. (2008). *Equipment Guide & Resources.* Retrieved 10 26, 2016, from Oregon Coalition for Health Care Ergonomics.

Garg, A. (1999). *Long-Term Effectiveness of "Zero-Lift Program" in Seven Nursing Homes and One Hospital.* Cincinnati, Ohio: National Institute for Occupational Safety and Health.

Ha, C., Cao, W., & Khasawneh, T. (2014). Ergonomic Assessment of Patient Under-arm Lifting Technique Using Digital Human Modeling. *Industrial and Systems Engineering Research Conference* (pp. 4066-4074). Montreal, Canada: Institute of Industrial Engineers.

Haong, K. H., & Mombaur, K. (2015). Adjustment to de Leva-anthropometric regression data for the changes in body proportions in elderly humans. *Journal of Biomechanics*, 3732-3736.

Hayman, K. J., Kerse, N., Dyall, L., Kepa, M., Teh, R., Wham, C., . . . Jatrana, S. (2012). *Life and Living in Advanced Age: A Cohort Study in New Zealand - Te Puawaitanga o Nga Tapuwae Kia Ora Tonu, LiLACS NZ: Study Protocol.* Auckland, New Zealand: BioMed Central Limited.

Holtermann, A., Clausen, T., Aust, B., Mortensen, O. S., & Andersen, L. L. (2013). Does occupational lifting and carrying among female health care workers contribute to an escalation of pain-day frequency? *European Journal of Pain*, 290-296.

Invacare Corporation. (2009, May). *Birdie 180*. Retrieved February 18, 2019, from Invacare New Zealand Website: https://www.invacare.co.nz/products/safe-patient-handling/lifters-and-accessories/birdie-180

Invacare Corporation. (2014). *Invacare Roze Stand Assist Lifter*. Retrieved February 19, 2019, from Invacare UK Website: http://www.invacare.co.uk/invacare-roze-stand-assist-lifter-ma-55rozen

Jusoh, M. A., Ismail, M. T., Rashid, H. B., Lokman, A. M., & Makhtar, A. K. (2018). Conceptual Design of the Mechanical Transfer Lift: Through the Application of EDP and KE. *Proceedings of the 7th International Conference on Kansei Engineering and Emotion Research 2018. KEER 2018. 739*, pp. 229-239. Springer, Singapore: Advances in Intelligent Systems and Computing.

Lachance, C. C., Korall, A. M., Russell, C. M., Feldman, F., Robinovitch, S. N., & Mackey, D. C. (2016). External Hand Forces Exerted by Long-Term Care Staff to Push Floor-Based Lifts: Effects of Flooring System and Resident Weight. *Human Factors: The Journal of the Human Factors and Ergonomics Society*, 927-943.

Lal, A., Moodie, M., Ashton, T., Siahpush, M., & Swinburn, B. (2012). Health care and lost productivity costs of overweight and obesity in New Zealand. *Australian and New Zealand Journal of Public Health*, 550-556.

Li, J., Wolf, L., & Evanoff, B. (2004). Use of mechanical patient lifts decreased musculoskeletal symptoms and injuries among health care workers. *Injury Prevention*, 212-216.

Marras, W. S., Knapik, G. G., & Ferguson, S. (2009). Lumbar spine forces during manoeuvring of ceiling-based and floor-based patient transfer devices. *Ergonomics 52:3*, 384-397.

Ministry of Health. (2002). *Health of Older People in New Zealand: A Statistical Reference.* Wellington, New Zealand: Ministry of Health.

Ministry of Health. (2015). *Understanding Excess Body Weight | New Zealand Health Survey.* Wellington: Ministry of Health.

Ministry of Social Development. (2014). *The New Zealand Carers' Strategy Action Plan for 2014 - 2018.* Wellington, New Zealand: Ministry of Social Development.

National Occupational Research Agency. (2009). *State of the Sector | Healthcare and Social Assistance.* Washington DC: United States of America: National Institute for Occupational Safety and Health.

Noble, N. L., & Sweeney, N. L. (2018, January). Barriers to the Use of Assistive Devices in Patient Handling. *Workplace Health and Safety, 66*(1), pp. 41-48.

North, N., Hughes, F., Rasmussen, E., Finlayson, M., Ashton, T., Campbell, T., & Tomkins, S. (2006). Use of Temporary Nurse Mechanisms by New Zealand's District Health Boards. *Labour, Employment and Work in New Zealand*, 278-286.

Office for Senior Citizens. (2008). *Highlights from the New Zealand Positive Ageing Report and Plan 2007 - 2010.* Wellington, New Zealand: Ministry of Social Development.

Office for Senior Citizens. (2015). *Business of Ageing.* Wellington, New Zealand: Ministry of Social Development.

Schoenfisch, A. L., Lipscomb, H. J., Pompeii, L. A., Myers, D. J., & Dement, J. M. (2013). Musculoskeletal injuries among hospital patient care staff before and after implementation of patient lift and transfer equipment. *Scandinavian Journal of Work, Enivornment, and Health*, 27-36.

Schoenfisch, A. L., Myers, D. J., Pompeii, L. A., & Lipscomb, H. J. (2011). Implementation and Adoption of Mechanical Patient Lift Equipment in the Hospital Setting: The Important of Organizational and Cultural Factors. *American Journal of Industrial Medicine*, 978-954.

Sorbye, L. W. (2009). *Frail homebound elderly: basic nurisng challenges of home care.* Tromso, Norway: University of Tromso.

Speser, S. (2011, April). Mechanical Lift Systems. *PN Magazine*, pp. 57-59.

Statistics New Zealand. (2007). *New Zealand's 65+ Population: A statistical volume.* Wellington, New Zealand: Statistics New Zealand.

Statistics New Zealand. (2009). *The Impact of Structural Population Change (Structural Change and the 65+ Population Articles).* Wellington, New Zealand: Statistics New Zealand.

Statistics New Zealand. (2014). *Disability Survey: 2013.* Wellington, New Zealand: Statistics New Zealand.

Tang, R., Holland, M., Milbauer, M., Olson, E., Skora, J., Kapellusch, J. M., & Garg, A. (2018). Biomechanical Evaluations of Bed-to-Wheelchair Transfer: Gait Belt Versus Walking Belt. *Workplace Health and Safety*, 384-392.

Tang, R., Poklar, M., Domke, H., Moore, S., Kapellusch, J., & Garg, A. (2016). Sit-To-Stand Lift: Effects of Lifted Height on Weight Borne and Upper Extremity Strength Requirements. *Research in Nursing and Health*, 9-14.

van der Woude, L. H., Geurts, C., Winkelman, H., & Veeger, H. (2003). Measurement of wheelchair rolling resistance with a handle bar push technique. *Journal of Medical Engineering and Technology*, 249-258.

Vaughan, J., Driver, J., Hall, E., & Race, E. (2014). A New Model for Successful Safe Handling Programs. *5th International Conference on Applied Human Factors and Ergonomics.* Krakow, Poland: Standford Risk Authority; Atlas Lift Tech.

Virmavirta, M., & Isolehto, J. (2014). Determining the location of the body's center of mass for different groups of physically active people. *Journal of Biomechanics*, 1909-1913.

Waymouth, A. D. (2014). *Low Effort Patient Handling Devices.* Christchurch, New Zealand: University of Canterbury.

Wiles, J. L., Allen, R. E., Palmer, A. J., Hayman, K. J., Keeling, S., & Kerse, N. (2009). Older people and their social spaces: A study of well-being and attachment to place in Aotearoa New Zealand. *Social Science & Medicine*, 664-671.

Wiles, J. W., Rolleston, A., Pillai, A., Broad, J., Teh, R., Gott, M., & Kerse, N. (2017). Attachment to place in advanced age: A study of the LiLACS. *Social Science and Medicine 185*, 27-37.

Wilson, N., Russell, D., & Wilson, B. (1993). *Size and shape of New Zealanders: New Zealand norms for anthropometric data.* Dunedin, New Zealand: Life in New Zealand Activity and Health Research Unit, University of Otago.

14 Appendices

APPENDIX A Scope and Methodology

The scope of this project is defined by the following research questions:

→ *Theoretically, how does the variation of patient height, weight, and distribution alter the carer input forces, and how does this compare to results from physical testing?*
→ *Theoretically, how does the configuration of kneepad, pivot point, and chest pad heights alter the carer input forces, and how does this compare to results from physical testing?*
→ *Theoretically, what is the lowest force want line and why?*
→ *In testing, what impact does a passive lift have on the carer input forces when compared with a live lift?*
→ *What testing techniques are suitable to assess carer input force?*
→ *How can the initial carer input force spike be removed or limited?*

The intended approach to answer these questions is detailed below.

Theoretically, how does the variation of patient height, weight, and distribution alter the carer input forces, and how does this compare to results from physical testing?

→ Development of MATLAB code A from Equations 1 – 14 detailed in Section 3.1 to calculate carer input force for a given lift path
→ MATLAB code B developed to calculate carer input force for a given lift path for varied total height and weight patients
→ MATLAB code C developed further to calculate carer input force for a given lift path for a variety of weight and height distributions
→ Development of Ply and Steel adjustable weight and height dummy
→ Assessment of dummy's suitability including durability and safety, detailing areas of improvement
→ Improvement of areas detailed; it is expected that this may contain issues similar to upgrading hip joints for stability, and developing a hoist attachment for easy manoeuvring
→ Dummy tested using methods detailed below
→ Dummy total height adjusted and tested using methods detailed below
→ Height adjustment test results compared with theoretical findings from MATLAB code B
→ Dummy total weight adjusted and tested using methods detailed below
→ Weight adjustment test results compared with theoretical findings for carer input force from MATLAB code B
→ Dummy segment weights and lengths adjusted and measured
→ Segment adjustment test results compared with theoretical findings for carer input force from MATLAB code C
→ Produce summary on how to adjust lifter setup to provide lower forces
→ Provide recommendations on what lifter setup would be required for different heights and weights

Theoretically, how does the configuration of kneepad, pivot point, and chest pad heights alter the carer input forces, and how does this compare to results from physical testing?

→ Development of MATLAB code A from Equations 1 – 14 detailed in Section 3.1 to calculate carer input force for a given lift path
→ MATLAB code D developed to calculate carer input force for a given lift path for varied kneepad, pivot and chest pad height adjustments
→ Development and assessment of dummy as detailed above
→ Lifter kneepad adjusted and dummy test completed
→ Kneepad adjustment test results compared with theoretical findings for carer input force from MATLAB code D
→ Pivot height adjusted and dummy test completed
→ Pivot height adjustment test results compared with theoretical findings for carer input force from MATLAB code B
→ Provide feedback on possible lifter adjustment and effects

Theoretically, what is the lowest force want line and why?

→ From theory discussed in Section 3.1, develop understanding of where lowest force paths should lie
→ Develop written overview of the understanding of how lowest force want lines are produced
→ Development of MATLAB code A from Equations 1 – 14 detailed in Section 3.1 to calculate carer input force for a given lift path
→ MATLAB code E developed to calculate movement of the ALP to create zero force load path and generating the lowest force want line
→ Development and assessment of dummy as detailed above
→ Testing of dummy on lowest force line
→ Results assessed and MATLAB code updated to include overlooked values; it is anticipated this will include friction forces and carer handle weight
→ Use height, weight, and distribution tests and findings from above to develop an understanding of how these effect the lowest force want line
→ Use kneepad, chest pad, and pivot height tests and findings from above to develop an understanding of how these effect the lowest force want line

In testing, what impact does a passive lift have on the carer input forces when compared with a live lift?

→ Carer input force testing for a variety of live patients
→ For any patient heights and weights, simulate with dummy replica
→ Assess force profiles of live patients and dummy patients for similarities and differences; it is expected that there will be differences due to the dummy modelling only skeletal movement rather than muscular movement
→ MATLAB codes redeveloped to account for differences in dummy and live patients

What testing techniques are suitable to assess carer input force?

- → Assembly of prototype mechanical torque transducer; this device uses torsion in a shaft attached to lifter arm to provide a visual readout through an adapted spring scale device
- → Assessment of suitability; consideration to be given to repeatability, accuracy, and functionality
- → Comparison with basic spring balance measurements tangential to the ALP
- → Research and assessment of other device options
- → Consideration of suitability of these devices; if neither are suitable a torque transducer may be sourced
- → Consideration of what forces will be measured against; it is expected that forces should be measured against the horizontal distance of the ALP from a datum point, possibly the pivot position

How can the initial carer input force spike be removed or limited?

- → Currently, this may have been resolved through bringing the torso of the patient to 20 degrees before loading the patient. It will become apparent whether this is so during the live patient and dummy testing, as will ways to mitigate this

APPENDIX B ZERO FORCE MATLAB CODE

```
function [COMXX,COMYY,ALPX,ALPY,HF,ALPA,CA,TORSO,HIPA,SHy]=
L0145_FORCE_DRIVEN_LCM_FUNC(H,W,SF,seat,TSF,COM,ALP,PERSON,PERW,PERK,pivrad,FORCE,
SA)
% ZERO FORCE / FORCE DRIVEN FUNCTION: Used to calculate zero force trajectories
%   Uses input of Height, Weight, Scale Factor, Seat Height, Thigh Length Percent
of Total Height, LCM Horizontal Adjustment, ALP Horizontal Adjustment, Person
Number (defines height and weight distribution), Percentage of Total Weight Lifted,
Shank Length Percent of Total Height, R1, Required Force Profile, Shank Angle (λ)
respectively
%   Provides outputs of LCM Horizontal Distance from origin, LCM Vertical Distance
from origin, ALP Horizontal Distance from origin, ALP Vertical Distance from
origin, Handle Force, ε, δ, α, β, and Shoulder Height from origin respectively
LW=W*PERW;                 %Lifted Mass (kg)
PHL=250;                   %Distance from Chest pad to Patient Handle (mm)
CHL=350;                   %Distance from Patient Handle to Carer Handle (mm)
HipA=0;                    %Initial Hip Angle from Horizontal (deg)
for i=1:28                 %For loop - step through hip angle
    HipA=(HipA+1);
    TEST(i)=1e2;           %Initial Value for test below
        if i== 1           %Assess initial hip angle, heel, foot and knee positions
        Hipy=seat;
        Heelx=PERK*H*sin(SA*pi/180);
        Heely=.0425*H;
        Toex=Heelx-220*SF;
        Kneex=0;
        Kneey=PERK*H*cos(SA*pi/180)+Heely;
        Hipx=sqrt((TSF*H)^2-(Kneey-Hipy)^2)+Kneex;
        Z=(Hipy-Kneey)/(TSF*H);
        HipA=(asin(Z))*180/pi;
    else
        Heelx=PERK*H*sin(SA*pi/180);
        Heely=.0425*H;
        Kneex=0;
        Kneey=PERK*H*cos(SA*pi/180)+Heely;
        Hipx=TSF*H*cos(((HipA)*pi/180))+Kneex;
        Hipy=Kneey+TSF*H*sin((HipA )*pi /180);
    end
```

```
    for j=1:10001           %For loop - step through torso angle
        B=(j-1)/200+20;     %Calculates torso angle

%Function below calculates patient LCM - shown in Appendix C
[ALPx,ALPy,COMx,COMy,SDx,SDy,CHX,CHY]=L0143_COM_FUNCTION_ADJ_HW(HipA,B,PHL,CHL,H,W,
SF,Hipx,Hipy,COM,ALP,seat,PERSON,PERW,j,SA,PERK);
%Begin calculation of forces, simulation of trajectories
        if i>2
            if j>1
%[From Eq 10]
              TR(i,j)=(LW*9.81*(ALPx-COMx))/(sqrt((ALPx-Hipx)^2+(ALPy-Hipy)^2));
%[From Eq 9]
             TA(i,j)=TR(i,j)/cos(((B+11-HipA)*pi/180));
%[From Eq 8]
             FYChest(i,j)=LW*9.81-TA(i,j)*sin((HipA)*pi/180);
%[From Eq 7]
             FXChest(i,j)=TA(i,j)*cos(HipA/180*pi);
%[From Eq 6]
             FShin(i,j)=((.1+.093+.029)*W*9.81+TA(i,j)*sin(HipA*pi/180))/
             cos(SA/180*pi);
%[From Eq [5]
             Fknee(i,j)=TA(i,j)*cos((HipA)*pi/180)+FShin(i,j)*sin(SA/180*pi);
%[From Eq 11]
             CHEST(i,j)=sqrt(FYChest(i,j)^2+FXChest(i,j)^2);
%[From Eq 12]
             CTA(i,j)=(atan(FXChest(i,j)/FYChest(i,j)))*180/pi ;
%[From Eq 13]
             theta=+CTA(i,j)-alpa(i,j);
%[From Eq 14]
             ff=CHEST(i,j)*sin(theta*pi/180);
%[From Eq 17]
             alpa(i,j)=(atan((ALPy-ALPY(i-1))/abs(ALPx-ALPX(i-1))))*180/pi;
%Testing loop - Angle θ is minimised to ensure zero force (can be altered for a
%constant force calculation)
                if abs(B-TORSO(i-1))<30
                    if abs(CTA(i,j)-alpa(i,j))<abs(TEST(i))
                        ALPX(i)=ALPx;
                        ALPY(i)=ALPy;
                        COMXX(i)=COMx;
                        COMYY(i)=COMy;
                        CX(i)=FXChest(i,j);
                        CY(i)=FYChest(i,j);
                        CF(i)=CHEST(i,j);
                        CA(i)=CTA(i,j) ;
                        ALPA(i)=atand((ALPY(i)-ALPY(i-1))/abs(ALPX(i)-ALPX(i-1)));
                        THETADIFF(i)=+CA(i)-ALPA(i);
                        SHx(i)=SDx;
                        SHy(i)=SDy;
                        FF(i)=CF(i)*sin(THETADIFF(i)*pi/180);
%[From Eq 28]
                        HF(i)=(pivrad/1000*FF(i));
                        TEST(i)=THETADIFF(i);
                        TORSO(i)=B;
                        HIPA(i)=HipA;
                        HX(i)=Hipx;
                        HY(i)=Hipy;
                    end
                end
            end
        else                    %For setup, where i<2 or j=1
            if B==20
            ALPX(i)=ALPx;
            ALPY(i)=ALPy;
            COMXX(i)=COMx;
            COMYY(i)=COMy;
            TR(i,j)=(LW*9.81*(ALPx-COMx))/(sqrt((ALPx-Hipx)^2+(ALPy-Hipy)^2));
            TA(i,j)=TR(i,j)/cos(((B+11-HipA)*pi /180));
           FShin(i,j)=((.1+.093+.029)*W*9.81+TA(i,j)*sin(HipA*pi/180))/
```

```
            cos(SA/180*pi);
            Fknee(i,j)=TA(i,j)*cos((HipA)*pi/180)+FShin(i,j)*sin(SA/180*pi);
            FXChest(i,j)=TA(i,j)*cos(HipA/180*pi);
             FYChest(i,j)=LW*9.81-TA(i,j)*sin((HipA)*pi/180);
             CHEST(i,j)=sqrt(FYChest(i,j)^2+FXChest(i,j)^2);
             CTA(i,j)=(atan(FXChest(i,j)/FYChest(i,j)))*180/pi ;
             CF(i)=CHEST(i,j);
             CA(i)=CTA(i,j) ;
             CA(i)=CTA(i,j) ;
             ALPA(i)=CA(i);
             THETADIFF(i)=+CA(i)-ALPA(i);
             SHx(i)=SDx;
             SHy(i)=SDy;
             TORSO(i)=0;
             FF(i)=CF(i)*sin(THETADIFF(i)*pi/180);
             HF(i)=(pivrad/1000*FF(i));
             HIPA(i)=HipA;
             HX(i)=Hipx;
             HY(i)=Hipy;
              end
        end
    end
end

%Can be used to tidy the first section of plots if necessary
% for Z=1:3
%     Y=4-Z;
%     ALPA(Y)=ALPA(Y+1)-(ALPA(Y+2)-ALPA(Y+1));
%     THETADIFF(Y)= THETADIFF(Y+1)-(THETADIFF(Y+2)-THETADIFF(Y+1));
%     ALPX(Y)=ALPX(Y+1)-(ALPX(Y+2)-ALPX(Y+1));
%     ALPY(Y)=ALPY(Y+1)-(ALPY(Y+2)-ALPY(Y+1));
%     COMXX(Y)=COMXX(Y+1)-(COMXX(Y+2)-COMXX(Y+1));
%     COMYY(Y)=COMYY(Y+1)-(COMYY(Y+2)-COMYY(Y+1));
%     CX(Y)=CX(Y+1)-(CX(Y+2)-CX(Y+1));
%     CY(Y)=CY(Y+1)-(CY(Y+2)-CY(Y+1));
%     CF(Y)=CF(Y+1)-(CF(Y+2)-CF(Y+1));
%     CA(Y)=CA(Y+1)-(CA(Y+2)-CA(Y+1));
%     HF(Y)=HF(Y+1)-(HF(Y+2)-HF(Y+1));
%
% end

%Plots first and last position of patient, LCM and ALP Trajectories, and ε and δ
for x=1:length(ALPX)
    y=x;
    if y==2|| y==length(ALPX)
[~,~,~,~,~,~,~,~]=L0143_COM_FUNCTION_ADJ_HW(HIPA(y),TORSO(y),PHL,CHL,H,W,SF,HX(y),H
Y(y),0,0,seat,PERSON,PERW,-1,SA,PERK);
    end
end
subplot(1,2,1)
hold on
plot(ALPX,ALPY)
plot(COMXX,COMYY)
title('Handle and Mechanism Trajectories')
xlabel('Distance (mm)')
ylabel('Distance (mm)')
subplot(1,2,2)
plot(TORSO,HF)
title('Handle Forces')
xlabel('Torso Angle (deg)')
ylabel('Force (N)')
figure
plot(TORSO,THETADIFF,'r',TORSO,ALPA,'b',TORSO,CA,'k')
title('Angle Comparisons')
xlabel('Torso Angle (deg)')
ylabel('Angle (deg)')
legend('Difference in Angles','Normal to Angle','Chestpad Angle')
end
```

APPENDIX C PATIENT CENTRE OF MASS MATLAB CODE

```
function[VLPx,VLPy,COMx,COMy,Shoulderx,Shouldery,CHX,CHY]=
L0143_DP_COM_FUNCTION_ADJ_HW(HipA,B,PHL,CHL,H,W,SF,Hipx,Hipy,COM,ALP,seat,
PERSON,PERW,j,SA,PERK,CPL,CPX,CPY)
% PATIENT CENTRE OF MASS CALCULATION: Used to calculate LCM and ALP positions
%   Uses input of β, α, Patient Handle Length, Carer Handle Length, Height, Weight,
Scale Factor, Horizontal Position of Hip from Origin, Vertical Position of Hip from
Origin, LCM Horizontal Adjustment, ALP Horizontal Adjustment, Seat Height, Person
Number (defines height and weight distribution), Percentage of Total Weight Lifted,
Graphing Style Indicator, Shank Angle (λ), Shank Length Percent of Total Height,
Chest Pad Centre distance from contact point, Chest Pad Horizontal Location from
Origin, Chest Pad Vertical Location From Origin respectively
%   Provides outputs of ALP/VLP Horizontal Distance from origin, ALP/VLP Vertical
Distance from origin, LCM Horizontal Distance from origin, LCM Vertical Distance
from origin, Shoulder Horizontal Distance from origin, and Shoulder Vertical
Distance from origin respectively
Heely=.0425*H;
Kneex=0;
Kneey=PERK*H*cos(SA*pi/180)+Heely;
Heelx=PERK*H*sin(SA*pi/180);
Toex=Heelx-220*SF;
Toey=0;
CHX=0;
```

Below is a sample of LCM Calculations Using the Coefficients from Table 14 for Patient N

```
if PERSON==N
    COMfootx=(Heelx-Toex)*C25;
    COMfooty=(Heely-Toey)*C25;
    MassFx=COMfootx/100*C8*W;
    MassFy=COMfooty/100*C8*W;
    COMkneex=(Kneex-Heelx)*C24+Heelx;
    COMkneey=(Kneey-Heely)*C24+Heely;
    MassKx=COMkneex*C7*W;
    MassKy=COMkneey*C7*W;
    COMhipx=(Hipx-Kneex)*C23 +Kneex;
    COMhipy=(Hipy-Kneey)*C23 +Kneey;
    MassThighx=COMhipx*C6*W;
    MassThighy=COMhipy*C6*W;
    Shoulderx=-C11*H*sin(B*pi/180)+Hipx;
    Shouldery=C11*H*cos(B*pi/180)+Hipy;
    COMSx=(Shoulderx-Hipx)*C19 +Hipx;
    COMSy=(Shouldery-Hipy)*C19+Hipy;
    MassTorsox=COMSx*C2*W;
    MassTorsoy=COMSy*C2*W;
    Elbowx=-C12*H*sin((-B+30)*pi/180)+Shoulderx;
    Elbowy=-C12*H*cos((-B+30)*pi/180)+Shouldery;
    COMEx=(Elbowx-Shoulderx)*C20 +Shoulderx;
    COMEy=(Elbowy-Shouldery)*C20+Shouldery;
    MassUAx=COMEx*C3*W;
    MassUAy=COMEy*C3*W;
    Wristx= Elbowx-C13*H*sin((B+30)*pi /180);
    Wristy=Elbowy+C13*H*cos((B+30)*pi /180);
    COMWx=(Wristx-Elbowx )*C21 +Elbowx ;
    COMWy=(Wristy-Elbowy )*C21+Elbowy ;
    MassFAx=COMWx*C4*W;
    MassFAy=COMWy*C4*W;
    Fingerx=Wristx-C14*H*sin((B+30)*pi /180);
    Fingery=Wristy+C14*H*cos((B+30)*pi /180);
    COMFx=(Fingerx-Wristx )*C22 +Wristx ;
    COMFy=(Fingery-Wristy )*C22 +Wristy ;
    MassHx=COMFx*C5*W;
    MassHy=COMFy*C5*W;
    Headx=-C10*H*sin(B*pi /180)+Shoulderx ;
    Heady=C10*H*cos(B*pi /180)+Shouldery ;
    COMHeadx=(Headx-Shoulderx )*C18 +Shoulderx ;
    COMHeady=(Heady-Shouldery )*C18+Shouldery ;
    MassHeadx=COMHeadx*C1*W;
```

```
    MassHeady=COMHeady*C1*W;
    COMx=(MassHeadx+MassHx*2+MassFAx*2+MassUAx*2+MassTorsox+MassThighx)/
(W*C9)-COM;
    COMy=(MassHeady+MassHy*2+MassFAy*2+MassUAy*2+MassTorsoy+MassThighy)/
(W*C9);
    CHY=CPL*C11*1800;
    VLPx=-CHY*sin((B)*pi /180)+Hipx-ALP-CPX*cosd(B) ;
    VLPy=CHY*cos((B)*pi /180)+Hipy-CPY*sind(B);
end
%________________________________________________________________________%

%Develop plot of patient
PERSONX=[Heelx,Toex,Heelx,Kneex,Hipx,Shoulderx,Elbowx,Wristx,Fingerx,
         Wristx,Elbowx,Shoulderx,Headx];
PERSONY=[Toey,Toey,Heely,Kneey,Hipy,Shouldery,Elbowy,Wristy,Fingery,
         Wristy,Elbowy,Shouldery,Heady];
if j==-1
        person=subplot(1,2,1);
        person=plot(PERSONX,PERSONY,'k--d');
        hold on
        person.LineWidth= 2;
end
if j==-2
        person=subplot(1,2,1);
        person=plot(PERSONX,PERSONY,'k-o');
        hold on
        person.LineWidth= 3;
end
end
```

APPENDIX D OVERVIEW MATLAB CODE

```
% OVERVIEW: Used to handle single pivot, tilting chestpad and force defined
function and set out all relevant variables

clc
clear all
close all
WEIGHT=[50;80;100];         %Defines weight in kg
HEIGHT=[1500;1800;1950];    %Defines height in mm
seat=480;                   %Defines seat height in mm
B=0;                        %Loop counter
COM=0;                      %LCM horizontal adjustment
ALP=0;                      %ALP horizontal adjustment
L=114;                      %R2 in mm
theta=25.9;                 %Angle of R2 from perpendicular to torso
HL=600;                     %Handle Length (patient handle to second carer handle)
SA=0; %21                   %Shank Angle in deg

%DEFINE PERSON NUMBER
% %   1   Standard Person, from de Leva 1996 (exrx.net)
% PERSON=1;
% TSF=.232;
% PERW=.7714;
% PERK=.247
% PIVstd=522;
% %   2   Sample Patient
PERSON=2;
TSF=.22
PERK=.247;
PERW=.778;
PIVstd=623
%   3   High LCM
% PERSON=3;
% TSF3=.225;
% PERK3=.247;
% PERW3=.777;
```

```
% PIVstd=498;
% %   4   Low LCM
% PERSON=4;
% TSF4=.225;
% PERK4=.247
% PERW4=.765;
% PIVstd=527;
%   5   Lower Body Dominant
% PERSON=5;
% TSF=.27;
% PERK=.3
% PERW=.7714
% PIVstd =642; check
%   6   Upper Body Dominant
% PERSON=6;
% PERK=.17
% TSF=.2;
% PERW=.7714
% PIVstd=987; check
%   7   Anna
% PERSON=7;
% PERK=.248
% TSF=.26;
% PERW=.778

%Alternate Pivot Point
PivLx1=-90;
PivLy1=35;
Pivrad1=680;

%Chest Pad Definitions
std=1;
largeblue=2;
bullhorns=3;
hug=4;
pink=5;
test=6;

CP=hug;                    %Defining Chest Pad Used

if CP==1;
    CPL=.8;                %CPL = ζ
    CPX=100;               %ALP/VLP X adjustment
    CPY=100;               %ALP/VLP Y adjustment
    CPW=2;                 %Chest Pad weight
end
if CP==2;
    CPL=.7;
    CPX=125;
    CPY=200;
    CPW=3;
end
if CP==3;
    CPL=.8;
    CPX=100;
    CPY=100;
    CPW=1.5;
end
if CP==4;
    CPL=.5;
    CPX=50;
    CPY=50;
    CPW=3;
end
if CP==5;
    CPL=.85;
    CPX=50;
    CPY=50;
```

```
    CPW=1;
end
if CP==6;
    CPL=0.55;
    CPX=00;
    CPY=00;
    CPW=3;
end

%Generating results for a number of heights and weights
for H=1:3
    SE=seat;
    HE=HEIGHT(H);
    SF=HE/1791;             %Scale Factor
    Hipy=seat;
    Pivstdx=224-61          %Horizontal Pivot Location
    Pivstdy=45              %Vertical Pivot Location
    for W=1:3
        B=B+1               %Loop Counter/Matrix Indexer
        WE=WEIGHT(W);
        %For tilting chest pad current
        [COMXSP(:,(B)),COMYSP(:,(B)),ALPXSP(:,(B)),ALPYSP(:,(B)),FORCESP(:,(B)),
        ALPASP(:,(B)),CASP(:,(B)),TORSOSP(:,B),HIPSP(:,(B))] =
        L0146_DP_PIVOT_SIMULATION_FUNC(HE,WE,SF,SE,TSF,COM,ALP,PERSON,PERW,PERK,
        PIVstd,Pivstdx,Pivstdy,40,18001,L,SA,theta,CPW,CPL,CPX,CPY);
        %For Zero Force
        [COMXZF(:,(B)),COMYZF(:,(B)),ALPXZF(:,(B)),ALPYZF(:,(B)),FORCEZF(:,(B)),

       ALPAZF(:,(B)),CAZF(:,(B)),TORSOZF,HIPAZF(:,(B))]=L0145_FORCE_DRIVEN_LCM_FUNC
        (HE,WE,SF,SE,TSF,0,0,PERSON,PERW,PERK,PIV,0,1,L,theta,PivLx1,PivLy1,L,HL);
        %Below: for alternate pivot point
        [COMXAP(:,(B)),COMYAP(:,(B)),ALPXAP(:,(B)),ALPYAP(:,(B)),FORCEAP(:,(B)),
        ALPAAP(:,(B)),CAAP(:,(B)),TORSOAP(:,B),HIPAP(:,(B))] =
        L0146_DP_PIVOT_SIMULATION_FUNC(HE,WE,SF,SE,TSF,COM,ALP,PERSON,PERW,PERK,
        pivrad1,PivLx1,PivLy1,30,10001,L,theta);
        PHL=350;     %Distance from Chest pad to Patient Handle (mm)
       CHL=320;     %Distance from Patient Hnadle to Carer Handle
       Hipy=seat;
       Heelx=220*SF;
       Heely=.0425*HE;
       Kneex=-PERK*HE*sin(21*pi/180)+Heelx;
       Kneey=PERK*HE*cos(21*pi/180)+Heely;
       if Kneey<seat
           Kneey=seat;
       end
       Hipx=sqrt((TSF*HE)^2-(Kneey-Hipy)^2)+Kneex;
       Z=(Kneey-Hipy)/(TSF*HE);
        HipA=(asin(Z))*180/pi;
        [ALPx2,ALPy2,COMx2,COMy2,T,T,T,T,T]=L0143_COM_FUNCTION_ADJ_HW
        (HIPAP(length(HIPAP)),TORSOAP(length(TORSOAP)),PHL,CHL,HE,WE,SF,
        Hipx, Hipy,0,0,SE,PERSON,PERW,-1,SA,PERK);
        figure
        subplot(1,2,1)
        hold on
        grid on
        plot(COMXSP,COMYSP,'b.', COMXZF,COMYZF, 'k.', COMXAP,COMYAP,'r.')
        plot(ALPXSP,ALPYSP,'bx', ALPXZF,ALPYZF, 'kx', ALPXAP,ALPYAP,'rx')
        legend('Patient','Patient Final Postion Adj Piv','Tilting Chestpad
         LCM','Zero Force LCM','Adjusted Pivot 1 LCM')
        title('Patient Lift Paths')
        xlabel('Distance (mm)')
        ylabel('Distance(mm)')
        subplot(1,2,2)
        plot(HIPSP,FORCESP,'b-+',HIPAZF,FORCEZF,'k-o', HIPAP,FORCEAP,'r.-')
        legend('Tilting Chestpad ','Zero Force','Adjusted Pivot
         1','Location','southwest')
        title('Handle Force')
        xlabel('Hip Angle (degrees)')
```

```
        ylabel('Force (N)')
      figure
    end
end
```

APPENDIX E SINGLE PIVOT MATLAB CODE

```
function [COMXX,COMYY,ALPX,ALPY,HF,ALPA,CA,TORSO,HIPA] =
L0146_PIVOT_SIMULATION_FUNC(H,W,SF,seat,TSF,COM,ALP,PERSON,PERW,PERK,pivrad,PivLx,P
ivLy,i,j,SA,TPW)
% SINGLE PIVOT: Used to calculate LCM and ALP trajectories and Handle Force
%   Uses input of Height, Weight, Scale Factor, Seat Height, Thigh Length Percent
of Total Height, LCM Horizontal Adjustment, ALP Horizontal Adjustment, Person
Number (defines height and weight distribution), Percentage of Total Weight Lifted,
Shank Length Percent of Total Height, R1, Pivot Horizontal Location From Origin,
Pivot Vertical Location from Origin, Number of Iterations for Hip, Number of
Iterations for Torso, Shank Angle (λ), and Thigh Weight Percent of Total Weight
respectively
%   Provides outputs of LCM Horizontal Distance from origin, LCM Vertical Distance
from origin, ALP Horizontal Distance from origin, ALP Vertical Distance from
origin, Handle Force, ε, δ, α, and β respectively

LW=W*PERW;                  %Lifted Mass (kg)
PHL=250;                    %Distance from Chest pad to Patient Handle (mm)
CHL=350;                    %Distance from Patient Handle to Carer Handle (mm)
HipA=0;                     %Initial Hip Angle from Horizontal (deg)
for i=1:i                   %For loop - step through hip angle
    HipA=(HipA+1);
    TEST1(i)=1e4;           %Initial Value for test below
  if i==1                   %Assess initial hip angle, foot and knee positions
        Hipy=seat;
        Heely=.0425*H;
        Kneex=0;
        Kneey=PERK*H*cos(SA*pi/180)+Heely;
        Hipx=sqrt((TSF*H)^2-(Kneey-Hipy)^2)+Kneex;
        Z=(Hipy-Kneey)/(TSF*H);
        HipA=(asin(Z))*180/pi;
    else
        Heelx=PERK*H*sin(SA*pi/180);
        Heely=.0425*H;
        Kneex=0;
        Kneey=PERK*H*cos(SA*pi/180)+Heely;
        Hipx=TSF*H*cos(((HipA)*pi/180))+Kneex;
        Hipy=Kneey+TSF*H*sin((HipA )*pi /180);
        if Hipy<seat
            Hipy==seat;
        end
    end
    for j=1:j               %For loop - step through torso angle
        B=(j-1)/200;        %Calculates torso angle

%Function below calculates patient LCM - shown in Appendix C
[ALPx,ALPy,COMx,COMy,SDx,SDy,CHX,CHY]=L0143_COM_FUNCTION_ADJ_HW(HipA,B,PHL,CHL,H,W,
SF,Hipx,Hipy,COM,ALP,seat,PERSON,PERW,j,SA,PERK);
%Begin calculation of forces, simulation of trajectories
        if j>1
            if i>2
%Testing: trajectory is acceptable when ALPy is closest to E in iteration
                E=sqrt((pivrad)^2-((ALPx-PivLx)^2))+PivLy;
                TEST(i,j)=E-ALPy;
                if abs(TEST(i,j))<abs(TEST1(i))
                    if abs(B-TORSO(i-1))<10
%[From Eq 10]
                    TR(i,j)=(LW*9.81*(ALPx-COMx))/(sqrt((ALPx-Hipx)^2+(ALPy-
                    Hipy)^2));
%[From Eq 9]
                    TA(i,j)=TR(i,j)/cos(((B+11-HipA)*pi/180));
```

```
%[From Eq 8]
                    FYChest(i,j)=LW*9.81-TA(i,j)*sin((HipA)*pi/180);
%[From Eq 7]
                    FXChest(i,j)=TA(i,j)*cos(HipA/180*pi);
%[From Eq 6]
                    FShin(i,j)=((.1+.093+.029)*W*9.81+TA(i,j)*sin(HipA*pi/180))/
                    cos(SA/180*pi);
%[From Eq [5]
                    Fknee(i,j)=TA(i,j)*cos((HipA)*pi/180)+FShin(i,j)*sin(SA/180*pi)
%[From Eq 11]
                    CHEST(i,j)=sqrt(FYChest(i,j)^2+FXChest(i,j)^2);
%[From Eq 12]
                    CTA(i,j)=(atan(FXChest(i,j)/FYChest(i,j)))*180/pi ;
                    ALPX(i)=ALPx;
                    ALPY(i)=ALPy;
                    COMXX(i)=COMx;
                    COMYY(i)=COMy;
                    CX(i)=FXChest(i,j);
                    CY(i)=FYChest(i,j);
                    CF(i)=CHEST(i,j);
                    CA(i)=-CTA(i,j) ;
                    ALPA(i)=atand((ALPY(i)-ALPY(i-1))/abs(ALPX(i)-ALPX(i-1)));
                    THETADIFF(i)=CA(i)-ALPA(i);
                    SHx(i)=SDx;
                    SHy(i)=SDy;
                    FF(i)=CF(i)*sin(THETADIFF(i)*pi/180);
%Function below calculates Handle COM - shown in Appendix F
                   [R,RR(i),COMX(i)]=L0150_HANDLE_COM_FUNC(PivLx,PivLy,pivrad,B,
                    PHL,CHL,H);
                    HF(i)=-((pivrad*FF(i)-(HW*(-COMX(i)+PivLx)))/(pivrad+CHL));
                    TORSO(i)=B;
                    HIPY(i)=Hipy;
                    HIPA(i)=HipA;
                    end
                end
            end
        else                %For setup, where i<2 or j=1
            ALPX(i)=ALPx;
            ALPY(i)=ALPy;
            COMXX(i)=COMx;
            COMYY(i)=COMy;
            TR(i,j)=(LW*9.81*(ALPx-COMx))/(sqrt((ALPx-Hipx)^2+(ALPy-Hipy)^2));
            TA(i,j)=TR(i,j)/cos(((B+11-HipA)*pi /180));
            FShin(i,j)=((.1+.093+.029)*W*9.81+TA(i,j)*sin(HipA*pi/180))/
            cos(SA/180*pi);
            Fknee(i,j)=TA(i,j)*cos((HipA)*pi/180)+FShin(i,j)*sin(SA/180*pi);
            FXChest(i,j)=TA(i,j)*cos(HipA/180*pi);
            FYChest(i,j)=LW*9.81-TA(i,j)*sin((HipA)*pi/180);
            CHEST(i,j)=sqrt(FYChest(i,j)^2+FXChest(i,j)^2);
            CTA(i,j)=(atan(FXChest(i,j)/FYChest(i,j)))*180/pi ;
            CF(i)=CHEST(i,j);
            CA(i)=CTA(i,j) ;
            CA(i)=CTA(i,j) ;
            ALPA(i)=CA(i);
            THETADIFF(i)=+CA(i)-ALPA(i);
            SHx(i)=SDx;
            SHy(i)=SDy;
            FF(i)=CF(i)*sin(THETADIFF(i)*pi/180);
            TORSO(i)=0;
            [R,RR(i)] = L0150_HANDLE_COM_FUNC(PivLx,PivLy,pivrad,B,PHL,CHL,H);
            HF(i)=-((pivrad*FF(i)+(HW*R*sin(((B-(20-
            ha))*pi()/180))))/(pivrad+CHL));
            HIPA(i)=HipA;
            HX(i)=Hipx;
            HY(i)=Hipy;
            end
    end
end
```

```
%Can be used to tidy the first section of plots if necessary and add pivot location
to ALP trajectory
%for Z=1:6
%     Y=7-Z;
%      ALPA(Y)=ALPA(Y+1)-(ALPA(Y+2)-ALPA(Y+1));
%      THETADIFF(Y)= THETADIFF(Y+1)-(THETADIFF(Y+2)-THETADIFF(Y+1));
%      ALPX(Y)=ALPX(Y+1)-(ALPX(Y+2)-ALPX(Y+1));
%      ALPY(Y)=ALPY(Y+1)-(ALPY(Y+2)-ALPY(Y+1));
%      COMXX(Y)=COMXX(Y+1)-(COMXX(Y+2)-COMXX(Y+1));
%      COMYY(Y)=COMYY(Y+1)-(COMYY(Y+2)-COMYY(Y+1));
%      COMYZ(Y)=COMYZ(Y+1)-(COMYZ(Y+2)-COMYZ(Y+1));
%      CX(Y)=CX(Y+1)-(CX(Y+2)-CX(Y+1));
%      CY(Y)=CY(Y+1)-(CY(Y+2)-CY(Y+1));
%      CF(Y)=CF(Y+1)-(CF(Y+2)-CF(Y+1));
%      CA(Y)=CA(Y+1)-(CA(Y+2)-CA(Y+1));
%      SHx(Y)=SHx(Y+1)-(SHx(Y+2)-SHx(Y+1));
%      SHy(Y)=SHy(Y+1)-(SHy(Y+2)-SHy(Y+1));
%      HF(Y)=HF(Y+1)-(HF(Y+2)-HF(Y+1));
%     if Y==1
%         ALPX(Y)=PivLx;
%         ALPY(Y)=PivLy;
%     end
%end

%Plots first and last position of patient, LCM and ALP Trajectories, and ε and δ
for x=1:length(ALPX)
    y=x;
    [~,~,~] = L0151_DP_HANDLE_COM_FUNC(PivLx,PivLy,pivrad,L,TORSO(y),PHL,
    CHL,ANG(y),HL,ALPX(y),ALPY(y),1);
    if y==3%|| y==length(ALPX)
    [~,~,~,~,~,~,~,~]=L0143_DP_COM_FUNCTION_ADJ_HW(HIPA(y),TORSO(y),PHL,
    CHL,H,W,SF,HX(y),HY(y),0,0,seat,PERSON,PERW,-2,SA,PERK,CPL,CPX,CPY)
    plot(COMXX,COMYY,'k*-')
    plot(ALPX,ALPY,'b.-')
    end
end

%Plots Free Body Diagram forces at a defined point
quiver(HX(10),HY(10),109*cosd(6.72),109*sind(6.71),'k')
quiver(HX(10),HY(10),114*cosd(.67),114*sind(.67),'k')
quiver(ALPX(10),ALPY(10),-113,0,'k')
quiver(ALPX(10),ALPY(10),0,518,'k')
quiver(Kneex,Kneey,-179.6*sind(21),179.6*cosd(21),'k')
quiver(Kneex-177.3,Kneey,177.3,0,'k')
quiver(0,10,0,167.7,'k')
quiver(0,10,64.3,0,'k')
```

APPENDIX F HANDLE CENTRE OF MASS MATLAB CODE

```
function [Radius,COMX,W,R1x,R1y]=L0151_DP_HANDLE_COM_FUNC_test(PivX,PivY,pivrad,
L,B,PHL,CHL,theta,HL,ALPx,ALPy,Y,CPW)
% TILTING CHESTPAD / DOUBLE PIVOT: Calculates LCM & ALP trajectories & Handle Force
%   Uses input of Pivot Horizontal Location From Origin, Pivot Vertical Location
from Origin, R2, β, Patient Handle Length, Carer Handle Length, ALP Horizontal
Distance from origin, ALP Vertical Distance from origin, Graphing Style Indicator,
and Chest Pad Weight respectively
%   Provides outputs of COM Radius from Pivot Point, Horizontal Location of Handle
COM from Origin, Handle Weight, Horizontal Location of Carer Handle from Origin,
and Vertical Location of Carer Handle from Origin respectively
INT=1;
if theta<0                  %Ensures angle is orientated correctly
    theta=-theta;
    INT=-1;
end

%R1
```

```
P2x=ALPx;
P2y=ALPy;
COMP1x=(P2x-PivX)*.5+PivX;
COMP1y=(P2y-PivY)*.5+PivY;
COMX1=pivrad*.002743*COMP1x;
COMY1=pivrad*.002743*COMP1y;

%R2
CPPx=INT*L*cos((theta)*pi()/180)+P2x;
CPPy=L*sin((theta)*pi()/180)+P2y;
COMCPx=(CPPx-P2x)*.1+P2x;
COMCPy=(CPPy-P2y)*.1+P2y;
COMX2=(L*.002743+2)*COMCPx;
COMY2=(L*.002743+2)*COMCPy;

%Patient Handle
PHP1x=CPPx-(L/4)*cos((theta)*pi()/180); %base point of handle
PHP1y=CPPy-(L/4)*sin((theta)*pi()/180); %base point of handle
PHP2x=-HL*sin((B)*pi/180)+PHP1x;
PHP2y=HL*cos((B)*pi/180)+PHP1y;
COMHx=(PHP2x-PHP1x)*.5+PHP1x;
COMHy=(PHP2y-PHP1y)*.5+PHP1y;
COMX3=(HL*.002743)*COMHx;
COMY3=HL*.002743*COMHy;

%Carer Handle 1
PHL1x=PHP1x-(HL/2)*sin((B)*pi()/180); %base point of handle
PHL1y=PHP1y+(HL/2)*cos((B)*pi()/180); %base point of handle
PHL2x=-PHL*cos((B)*pi/180)+PHL1x;
PHL2y=-PHL*sin((B)*pi/180)+PHL1y;
COMPHx=(PHL2x-PHL1x)*.5+PHL1x;
COMPHy=(PHL2y-PHL1y)*.5+PHL1y;
COMX4=(PHL*.002743)*COMPHx;
COMY4=PHL*.002743*COMPHy;

%Carer Handle 2
CHL2x=CHL*cos((B)*pi/180)+PHP2x;
CHL2y=CHL*sin((B)*pi/180)+PHP2y;
COMCHx=(CHL2x-PHP2x)*.5+PHP2x;
COMCHy=(CHL2y-PHP2y)*.5+PHP2y;
COMX5=(CHL*.002743)*COMCHx;
COMY5=CHL*.002743*COMCHy;

%Handle Weight
W=((pivrad+L+HL+PHL+CHL)*.002743+CPW);
COMX=(COMX1+COMX2+COMX3+COMX4+COMX5)/W;
COMY=(COMY1+COMY2+COMY3+COMY4+COMY5)/W;
W=W*9.81;
Radius=sqrt((COMX-PivX)^2+(COMY-PivY)^2);
Delta=atan((COMX-PivX)/(COMY-PivY))*180/pi;

%Plot Handle
if Y==1
A=[PivX,P2x,CPPx,PHP1x,PHP2x,PHL1x,PHL2x,PHL1x,PHP2x,CHL2x];
C=[PivY,P2y,CPPy,PHP1y,PHP2y,PHL1y,PHL2y,PHL1y,PHP2y,CHL2y];
plot(A,C)
hold on
end

%Carer Handle Position
R1x=PHP2x;
R1y=PHP2y;
end
```

APPENDIX G TILTING CHEST PAD MATLAB CODE

```
function [COMXX,COMYY,ALPX,ALPY,HF,ALPA,CA,TORSO,HIPA]=L0146_PIVOT_SIMULATION_FUNC
(H,W,SF,seat,TSF,COM,ALP,PERSON,PERW,PERK,pivrad,PivLx,PivLy,I1,j,L,SA,thet,
CPW,CPL,CPX,CPY)
% TILTING CHESTPAD / DOUBLE PIVOT: Calculates LCM & ALP trajectories & Handle Force
%   Uses input of Height, Weight, Scale Factor, Seat Height, Thigh Length Percent
of Total Height, LCM Horizontal Adjustment, ALP Horizontal Adjustment, Person
Number (defines height and weight distribution), Percentage of Total Weight Lifted,
Shank Length Percent of Total Height, R1, Pivot Horizontal Location From Origin,
Pivot Vertical Location from Origin, Number of Iterations for Hip, Number of
Iterations for Torso, R2, Shank Angle (λ), Initial Angle of Chest Pad from
Perpendicular to Chest, Chest Pad Weight, Chest Pad Centre distance from contact
point, Chest Pad Horizontal Location from Origin, and Chest Pad Vertical Location
From Origin respectively
%   Provides outputs of LCM Horizontal Distance from origin, LCM Vertical Distance
from origin, ALP Horizontal Distance from origin, ALP Vertical Distance from
origin, Handle Force, ε, δ, α, and β respectively

LW=W*PERW;                  %Lifted Mass (kg)
PHL=350;                    %Distance from Chest pad to Patient Handle (mm)
CHL=320;                    %Distance from Patient Handle to Carer Handle (mm)
HL=600;                     %Handle Length (mm)
HipA=0;                     %Initial Hip Angle from Horizontal (deg)
C=0-(1800-H)/30*2;          %Initial Torso Angle Adjustment with Height (deg)
for I=1:I1                  %For loop - step through hip angle
    HipA=(HipA+1);
    TEST1(I)=1e4;           %Initial Value for test below
   if I == 1                %Assess initial hip angle, foot and knee positions
        Hipy=seat;
        Heely=.0425*H;
        Kneex=0;
        Kneey=PERK*H*cos(SA*pi/180)+Heely;
        Hipx=sqrt((TSF*H)^2-(Kneey-Hipy)^2)+Kneex;
        Z=(Hipy-Kneey)/(TSF*H);
        HipA=(asin(Z))*180/pi;
    else
        Heelx=PERK*H*sin(SA*pi/180);
        Heely=.0425*H;
        Kneex=0;
        Kneey=PERK*H*cos(SA*pi/180)+Heely;
        Hipx=TSF*H*cos(((HipA)*pi/180))+Kneex;
        Hipy=Kneey+TSF*H*sin((HipA )*pi /180);
        SA=SA+16.5/40;              %Shank Angle Adjustment for Tilting Kneepads
        PivLx(I)=PivLx(I-1);        %Pivot Position Adjustment for Tilted Kneepads
    end
    for j=1:j                       %For loop - step through torso angle
        B=(j-1)/200+C;              %Calculates torso angle

%Function below calculates patient LCM - shown in Appendix C
[VLPx,VLPy,COMx,COMy,SDx,SDy,CHX,CHY]=L0143_DP_COM_FUNCTION_ADJ_HW(HipA,B,PHL,CHL,H
,W,SF,Hipx, Hipy,COM,ALP,seat,PERSON,PERW,j,SA,PERK,CPL,CPX,CPY);
%Begin calculation of forces, simulation of trajectories
        if j>1
            if I>2
               F=VLPx-L*cos((thet+B)/180*pi); %Calculated Value of ALPx (Figure 51)
               G=VLPy-L*sin((thet+B)/180*pi); %Calculated Value of ALPy (Figure 51)
                 RAD(I,j)=sqrt((PivLx(I)-F)^2+(PivLy-G)^2);
                if B-TORSO(I-1)<10
%Testing: Trajectory is acceptable when R1c is equal to R1 (Figure 51)
                 TEST(I,j)=abs(pivrad-RAD(I,j));
                if TEST(I,j)<TEST1(I)
                    ALPx(I,j)=F;
                    ALPy(I,j)=G;
                    M(I,j)=LW*9.81*(COMx-ALPx(I,j))/1000;
%[From Eq 10]
                    TR(I,j)=(M(I,j))/(CHY/1000);
%[From Eq 9]
```

```
                          TA(I,j)=TR(I,j)/cos(((B+11-HipA)*pi/180));
%[From Eq 8]
                          FYChest(I,j)=LW*9.81-TA(I,j)*sin((HipA)*pi/180);
%[From Eq 7]
                          FXChest(I,j)=TA(I,j)*cos(HipA/180*pi);
%[From Eq 6]
                          FShin(I,j)=((.1+.093+.029)*W*9.81+TA(I,j)*sin(HipA*pi/180))/
                          cos(SA/180*pi);
%[From Eq [5]
                          Fknee(I,j)=TA(I,j)*cos((HipA)*pi/180)+FShin(I,j)*sin(SA/180*pi)
%[From Eq 11]
                          CHEST(I,j)=sqrt(FYChest(I,j)^2+FXChest(I,j)^2);
%[From Eq 12]
                          CTA(I,j)=(atan(FXChest(I,j)/FYChest(i,j)))*180/pi ;
                          ALPX(I)=ALPx(I,j);
                          ALPY(I)=ALPy(I,j);
                          VLPX(I)=VLPx;
                          VLPY(I)=VLPy;
                          COMXX(I)=COMx;
                          COMYY(I)=COMy;
                          CX(I)=FXChest(I,j);
                          CY(I)=FYChest(I,j);
                          CF(I)=CHEST(I,j);
                          CA(I)=CTA(I,j)  ;
                          ALPA(I)=(atan((ALPY(I-1)-ALPY(I))/(ALPX(I-1)-ALPX(I))))*180/pi;
                          MM(I)=M(I,j);
                          THETADIFF(I)=+CA(I)-ALPA(I);
                          FF(I)=CF(I)*sin(THETADIFF(I)*pi/180);
%Function below calculates Handle COM - shown in Appendix F
                          [R(I),HCOMX,HW,R1x(I),R1y(I)]=L0151_DP_HANDLE_COM_FUNC_test
                          (PivLx(I),PivLy,pivrad,L,B,PHL,CHL,thet,HL,ALPx(I,j),ALPy(I,j),
                          0,CPW);
                           HFM(I)=FF(I)*pivrad/1000;
                           HF(I)=((HFM(I)/(sqrt((PivLx(I)-R1x(I))^2+(PivLy-R1y(I))^2)
                          /1000))*cosd(B)-HW*HCOMX/1000);
                          TORSO(I)=B;
                          HIPY(I)=Hipy;
                          HIPX(I)=Hipx;
                          HIPA(I)=HipA;
                      end
                  end
              end
          else                         %For setup, where i<2 or j=1
              ALPX(I)=320*SF;
              ALPY(I)=9*SF;
              COMXX(I)=COMx;
              COMYY(I)=COMy;
              TORSO(I)=20;
          end
      end
end

%Can be used to tidy the first section of plots if necessary and add pivot location
to ALP trajectory
% for Z=1:6
%     Y=7-Z;
%     ALPA(Y)=ALPA(Y+1)-(ALPA(Y+2)-ALPA(Y+1));
%     THETADIFF(Y)= THETADIFF(Y+1)-(THETADIFF(Y+2)-THETADIFF(Y+1));
%     ALPX(Y)=ALPX(Y+1)-(ALPX(Y+2)-ALPX(Y+1));
%     ALPY(Y)=ALPY(Y+1)-(ALPY(Y+2)-ALPY(Y+1));
%     COMXX(Y)=COMXX(Y+1)-(COMXX(Y+2)-COMXX(Y+1));
%     COMYY(Y)=COMYY(Y+1)-(COMYY(Y+2)-COMYY(Y+1));
%     THETADIFF(Y)=THETADIFF(Y+1)-(THETADIFF(Y+2)-THETADIFF(Y+1));
%     ALPA(Y)=ALPA(Y+1)-(ALPA(Y+2)-ALPA(Y+1));
%     CY(Y)=CY(Y+1)-(CY(Y+2)-CY(Y+1));
%     CF(Y)=CF(Y+1)-(CF(Y+2)-CF(Y+1));
%     CA(Y)=CA(Y+1)-(CA(Y+2)-CA(Y+1));
%     SHx(Y)=SHx(Y+1)-(SHx(Y+2)-SHx(Y+1));
```

```
%      SHy(Y)=SHy(Y+1)-(SHy(Y+2)-SHy(Y+1));
%      HF(Y)=HF(Y+1)-(HF(Y+2)-HF(Y+1));
%      if Y==1
%          ALPX(Y)=PivLx(1);
%          ALPY(Y)=PivLy;
%      end
% end

%Plots first and last positions of patient, and LCM and ALP trajectories
for x=1:length(ALPX)-1
    y=x+1;
    [~,~,~]=L0151_DP_HANDLE_COM_FUNC_test(PivLx(y),PivLy,pivrad,L,TORSO(y),PHL,
    CHL,thet+TORSO(y),HL,ALPX(y),ALPY(y),1,CPW);
 if y==3|| y==length(ALPX)
    [~,~,~,~,~,~,~,~]=L0143_DP_COM_FUNCTION_ADJ_HW(HIPA(y),TORSO(y),PHL,CHL,H,W,SF,
    HIPX(y), HIPY(y),0,0,seat,PERSON,PERW,-1,SA,PERK,CPL,CPX,CPY);
    plot(COMXX,COMYY,'k*-')
    plot(ALPX,ALPY,'b.-')
    end
end
end
```

APPENDIX H HTS3 MATLAB CODE

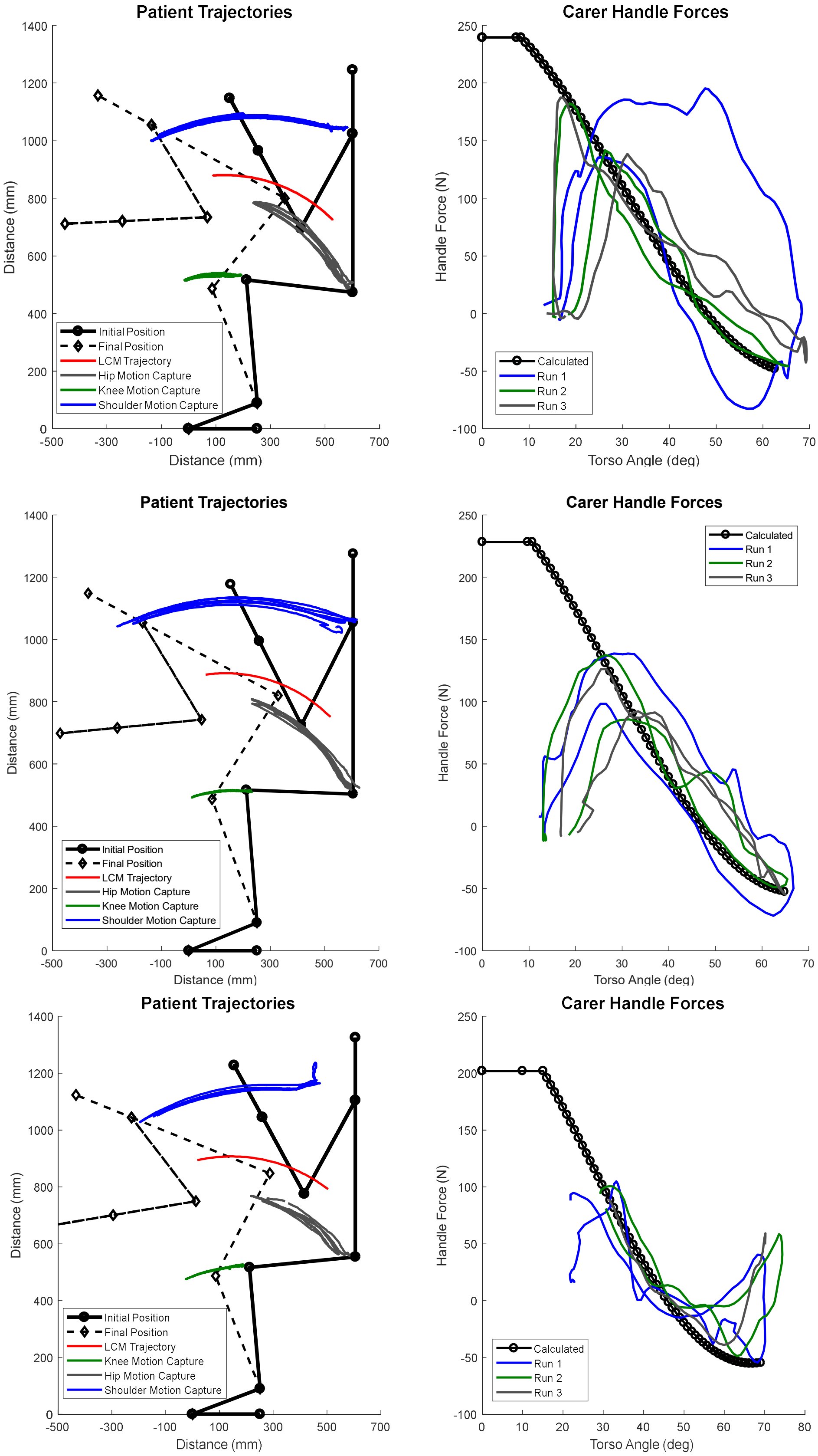
```
function [COMXX,COMYY,ALPX,ALPY,HF,ALPA,CA,TORSO,HIPA]=L0146_PIVOT_SIMULATION_FUNC
(H,W,SF,seat,TSF,COM,ALP,PERSON,PERW,PERK,pivrad,PivLx,PivLy,I1,j,L,SA1,thet,
CPW,CPL,CPX,CPY)
% ALTERED TILTING CHESTPAD / HTS3: Calculates LCM & ALP trajectories & Handle Force
%   Uses input of Height, Weight, Scale Factor, Seat Height, Thigh Length Percent
of Total Height, LCM Horizontal Adjustment, ALP Horizontal Adjustment, Person
Number (defines height and weight distribution), Percentage of Total Weight Lifted,
Shank Length Percent of Total Height, R1, Pivot Horizontal Location From Origin,
Pivot Vertical Location from Origin, Number of Iterations for Hip, Number of
Iterations for Torso, R2, Shank Angle (λ), Initial Angle of Chest Pad from
Perpendicular to Chest, Chest Pad Weight, Chest Pad Centre distance from contact
point, Chest Pad Horizontal Location from Origin, and Chest Pad Vertical Location
From Origin respectively
%   Provides outputs of LCM Horizontal Distance from origin, LCM Vertical Distance
from origin, ALP Horizontal Distance from origin, ALP Vertical Distance from
origin, Handle Force, ε, δ, α, and β respectively

LW=W*PERW;                  %Lifted Mass (kg)
PHL=475;                    %Distance from Chest pad to Patient Handle (mm)
CHL=475;                    %Distance from Patient Handle to Carer Handle (mm)
HL=475;                     %Handle Length (mm)
HipA=0;                     %Initial Hip Angle from Horizontal (deg)
C=0-(1800-H)/30*2;          %Initial Torso Angle Adjustment with Height (deg)
for I=1:I1                  %For loop - step through hip angle
    HipA=(HipA+1);
    TEST1(I)=1e4;           %Initial Value for test below
   if I == 1                %Assess initial hip angle, foot and knee positions
        Hipy=seat;
        Heelx=250*SF;
        Heely=.05*H;
        Kneex=-PERK*H*sin(SA*pi/180)+Heelx;
        Kneey=PERK*H*cos(SA*pi/180)+Heely;
        Hipx=sqrt((TSF*H)^2-(Kneey-Hipy)^2)+Kneex;
        Z=(Hipy-Kneey)/(TSF*H);
        HipA=(asin(Z))*180/pi;
    else
        Heelx=250*SF;
        Heely=.05*H;
        Kneex=-PERK*H*sin(SA*pi/180)+Heelx;
        Kneey=PERK*H*cos(SA*pi/180)+Heely;
        Hipx=TSF*H*cos(((HipA)*pi/180))+Kneex;
        Hipy=Kneey+TSF*H*sin((HipA )*pi /180);
        SA=SA+11/I1;              %Shank Angle Adjustment for Tilting Kneepads
```

```
        PivLx(I)=PivLx(I-1);      %Pivot Position Adjustment for Tilted Kneepads
    end
    for j=1:j                     %For loop - step through torso angle
        B=(j-1)/200+C;            %Calculates torso angle

%Function below calculates patient LCM - shown in Appendix C
[VLPx,VLPy,COMx,COMy,SDx,SDy,CHX,CHY]=L0143_DP_COM_FUNCTION_ADJ_HW_adj(HipA,B,PHL,
CHL,H,W,SF,Hipx, Hipy,COM,ALP,seat,PERSON,PERW,j,SA,PERK,CPL,CPX,CPY);
%Begin calculation of forces, simulation of trajectories
        if j>1
            if I>2
                F=pivrad*sind(-SA)+PivLx(I);     %Calculated Value of ALPx from IPP
                F1=VLPx-L*cos((thet+B)/180*pi); %Calculated Value of ALPx from VLP
                G=pivrad*cosd(-SA)+PivLy;       %Calculated Value of ALPy from IPP
                G1=VLPy-L*sin((thet+B)/180*pi); %Calculated Value of ALPy from VLP
                RAD(I,j)=sqrt((F1-F)^2+(G1-G)^2);
                if B-TORSO(I-1)<10
%Testing: Trajectory is acceptable when R1c is equal to R1 (Figure 51)
                    TEST(I,j)=abs(RAD(I,j));
                    if TEST(I,j)<TEST1(I)
                        ALPx(I,j)=F;
                        ALPy(I,j)=G;
                        M(I,j)=LW*9.81*(COMx-ALPx(I,j))/1000;
%[From Eq 10]
                        TR(I,j)=(M(I,j))/(CHY/1000);
%[From Eq 9]
                        TA(I,j)=TR(I,j)/cos(((B+11-HipA)*pi/180));
%[From Eq 8]
                        FYChest(I,j)=LW*9.81-TA(I,j)*sin((HipA)*pi/180);
%[From Eq 7]
                        FXChest(I,j)=TA(I,j)*cos(HipA/180*pi);
%[From Eq 6]
                        FShin(I,j)=((.1+.093+.029)*W*9.81+TA(I,j)*sin(HipA*pi/180))
                        /cos(SA/180*pi);
%[From Eq [5]
                        Fknee(I,j)=TA(I,j)*cos((HipA)*pi/180)+FShin(I,j)*sin
                        (SA/180*pi);
%[From Eq 11]
                        CHEST(I,j)=sqrt(FYChest(I,j)^2+FXChest(I,j)^2);
%[From Eq 12]
                        CTA(I,j)=(atan(FXChest(I,j)/FYChest(i,j)))*180/pi ;
                        ALPX(I)=ALPx(I,j);
                        ALPY(I)=ALPy(I,j);
                        VLPX(I)=VLPx;
                        VLPY(I)=VLPy;
                        COMXX(I)=COMx;
                        COMYY(I)=COMy;
                        CX(I)=FXChest(I,j);
                        CY(I)=FYChest(I,j);
                        CF(I)=CHEST(I,j);
                        CA(I)=CTA(I,j) ;
                       ALPA(I)=(atan((ALPY(I-1)-ALPY(I))/(ALPX(I-1)-ALPX(I))))*180/pi;
                        MM(I)=M(I,j);
                        THETADIFF(I)=+CA(I)-ALPA(I);
                        FF(I)=CF(I)*sin(THETADIFF(I)*pi/180);
%Function below calculates Handle COM - shown in Appendix F
                        [R(I),HCOMX(I),HW,R1x(I),R1y(I)]=
                        L0151_DP_HANDLE_COM_FUNC_test(PivLx(I),PivLy,pivrad,L,B,
                        PHL,CHL,thet,HL,ALPx(I,j),ALPy(I,j),0,CPW);
                        HFM(I)=FF(I)*pivrad/1000;
                       HF(I)=((HFM(I)/(sqrt((PivLx(I)-R1x(I))^2+(PivLy-R1y(I))^2)
                        /1000))*cosd(B)-HW*HCOMX(I)/1000);
                        TORSO(I)=B;
                        HIPY(I)=Hipy;
                        HIPX(I)=Hipx;
                        HIPA(I)=HipA;
                        TEST1(I)=TEST(I,j);
                    end
```

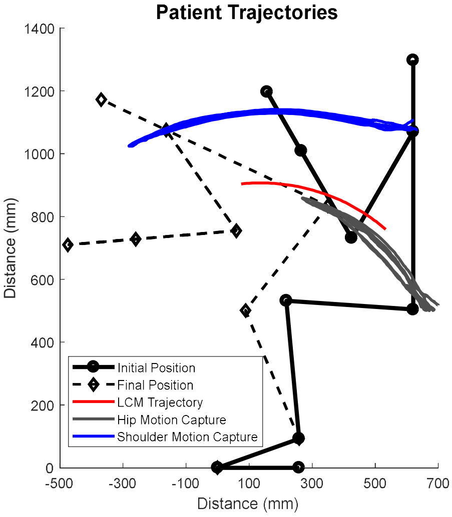
```
                end
            end
        else                %For setup, where i<2 or j=1
            ALPX(I)=320*SF;
            ALPY(I)=9*SF;
            COMXX(I)=COMx;
            COMYY(I)=COMy;
            TORSO(I)=0;
        end
    end
end
%Can be used to tidy the first section of plots if necessary and add pivot location
to ALP trajectory
% for Z=1:6
%     Y=7-Z;
%     ALPA(Y)=ALPA(Y+1)-(ALPA(Y+2)-ALPA(Y+1));
%     THETADIFF(Y)= THETADIFF(Y+1)-(THETADIFF(Y+2)-THETADIFF(Y+1));
%     ALPX(Y)=ALPX(Y+1)-(ALPX(Y+2)-ALPX(Y+1));
%     ALPY(Y)=ALPY(Y+1)-(ALPY(Y+2)-ALPY(Y+1));
%     COMXX(Y)=COMXX(Y+1)-(COMXX(Y+2)-COMXX(Y+1));
%     COMYY(Y)=COMYY(Y+1)-(COMYY(Y+2)-COMYY(Y+1));
%     THETADIFF(Y)=THETADIFF(Y+1)-(THETADIFF(Y+2)-THETADIFF(Y+1));
%     ALPA(Y)=ALPA(Y+1)-(ALPA(Y+2)-ALPA(Y+1));
%     CY(Y)=CY(Y+1)-(CY(Y+2)-CY(Y+1));
%     CF(Y)=CF(Y+1)-(CF(Y+2)-CF(Y+1));
%     CA(Y)=CA(Y+1)-(CA(Y+2)-CA(Y+1));
%     SHx(Y)=SHx(Y+1)-(SHx(Y+2)-SHx(Y+1));
%     SHy(Y)=SHy(Y+1)-(SHy(Y+2)-SHy(Y+1));
%     HF(Y)=HF(Y+1)-(HF(Y+2)-HF(Y+1));
%     if Y==1
%         ALPX(Y)=PivLx(1);
%         ALPY(Y)=PivLy;
%     end
% end

%Plots first and last positions of patient, and LCM and ALP trajectories
for x=1:length(ALPX)-1
    y=x+1;
    [~,~,~]=L0151_DP_HANDLE_COM_FUNC_test(PivLx(y),PivLy,pivrad,L,TORSO(y),PHL,
    CHL,thet+TORSO(y),HL,ALPX(y),ALPY(y),1,CPW);
 if y==3|| y==length(ALPX)
    [~,~,~,~,~,~,~,~]=L0143_DP_COM_FUNCTION_ADJ_HW(HIPA(y),TORSO(y),PHL,CHL,H,W,SF,
    HIPX(y), HIPY(y),0,0,seat,PERSON,PERW,-1,SA,PERK,CPL,CPX,CPY);
    plot(COMXX,COMYY,'k*-')
    plot(ALPX,ALPY,'b.-')
    end
end
```

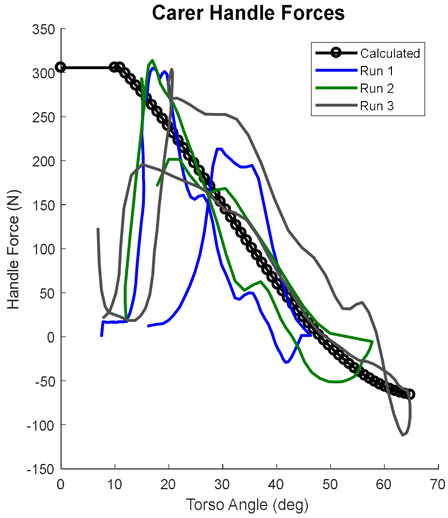
APPENDIX I HTS3 TEST RESULTS

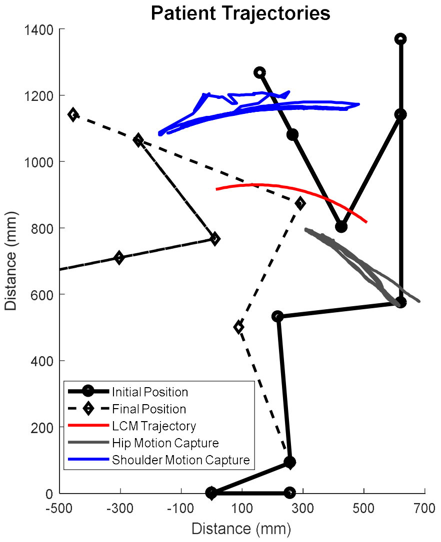
Patient A. At 460, 490 and 560 Millimetre Transfer Surfaces

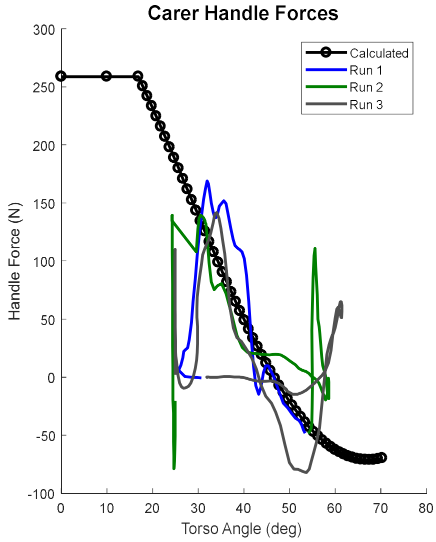


Patient B. At 490 and 560 Millimetre Transfer Surfaces

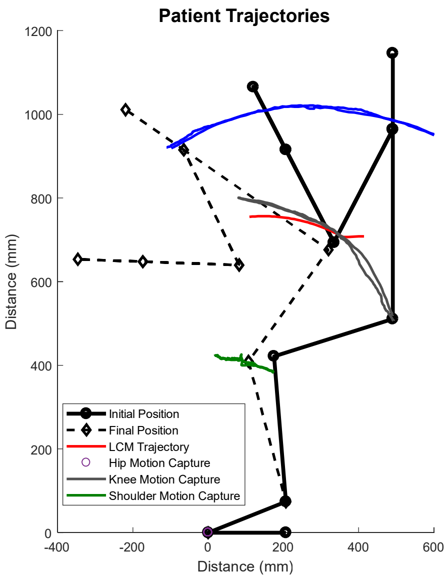


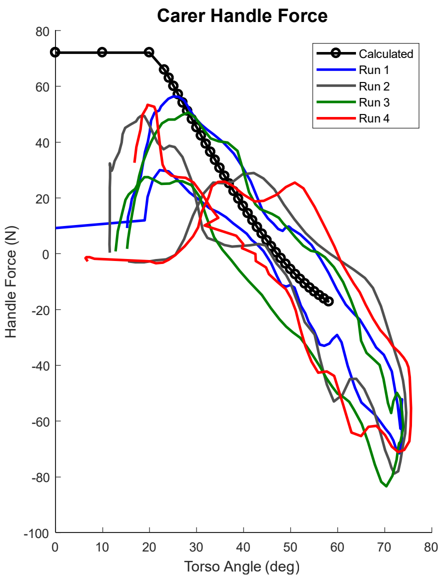




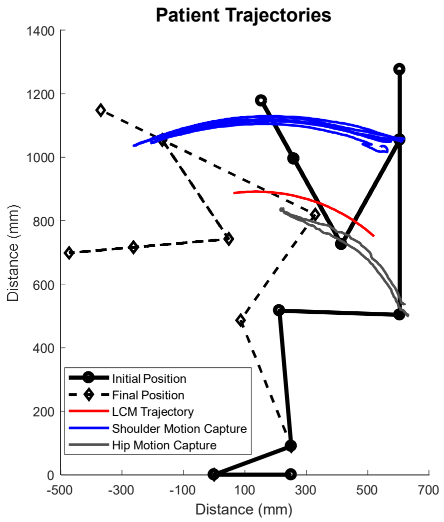


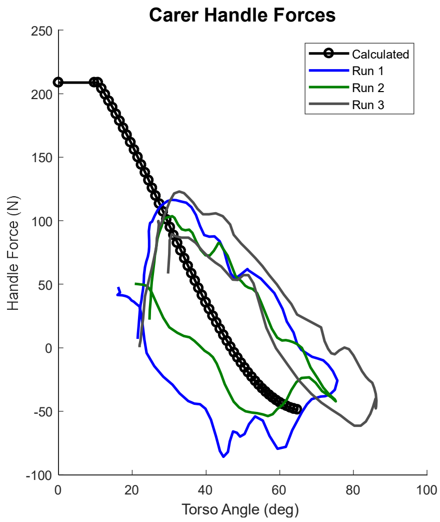
Patient C. At 490 Millimetre Transfer Surface



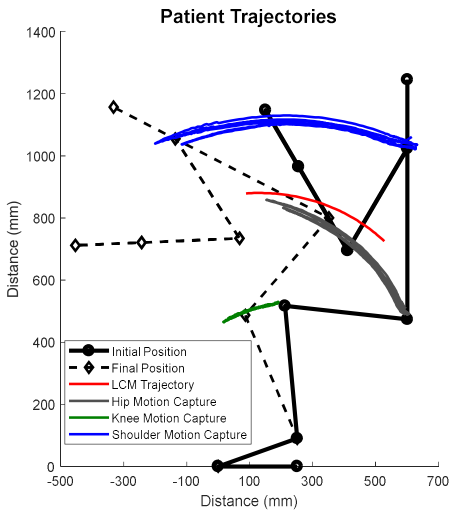


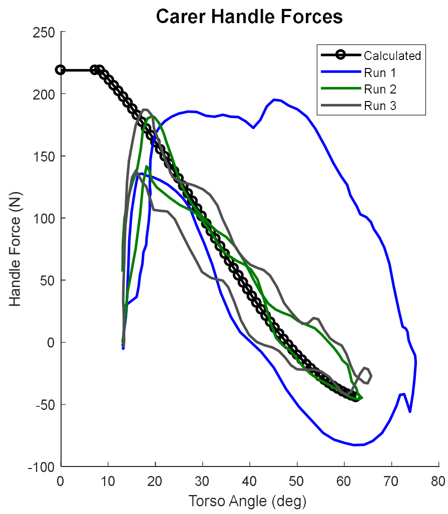
Patient D. At 490 Millimetre Transfer Surface

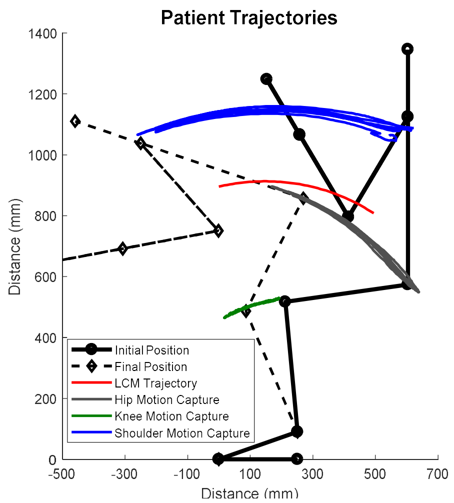


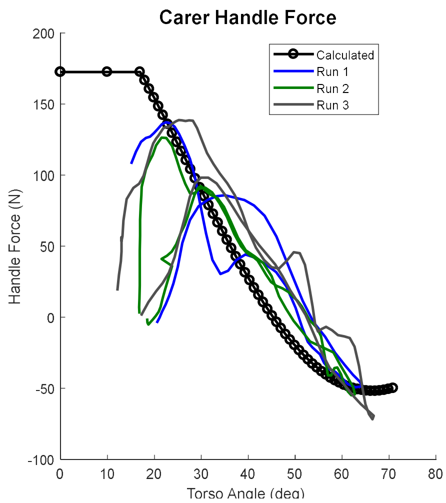


Patient E. At 460 and 560 Millimetre Transfer Surface

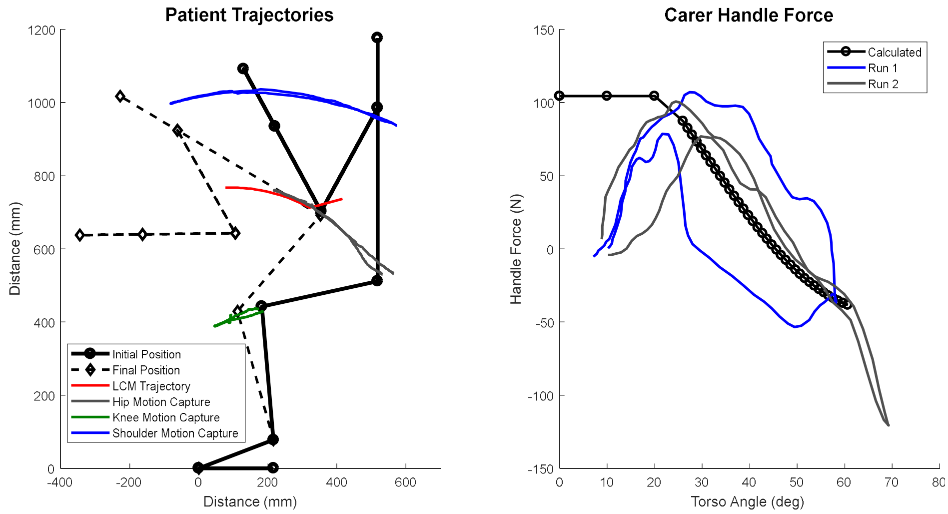




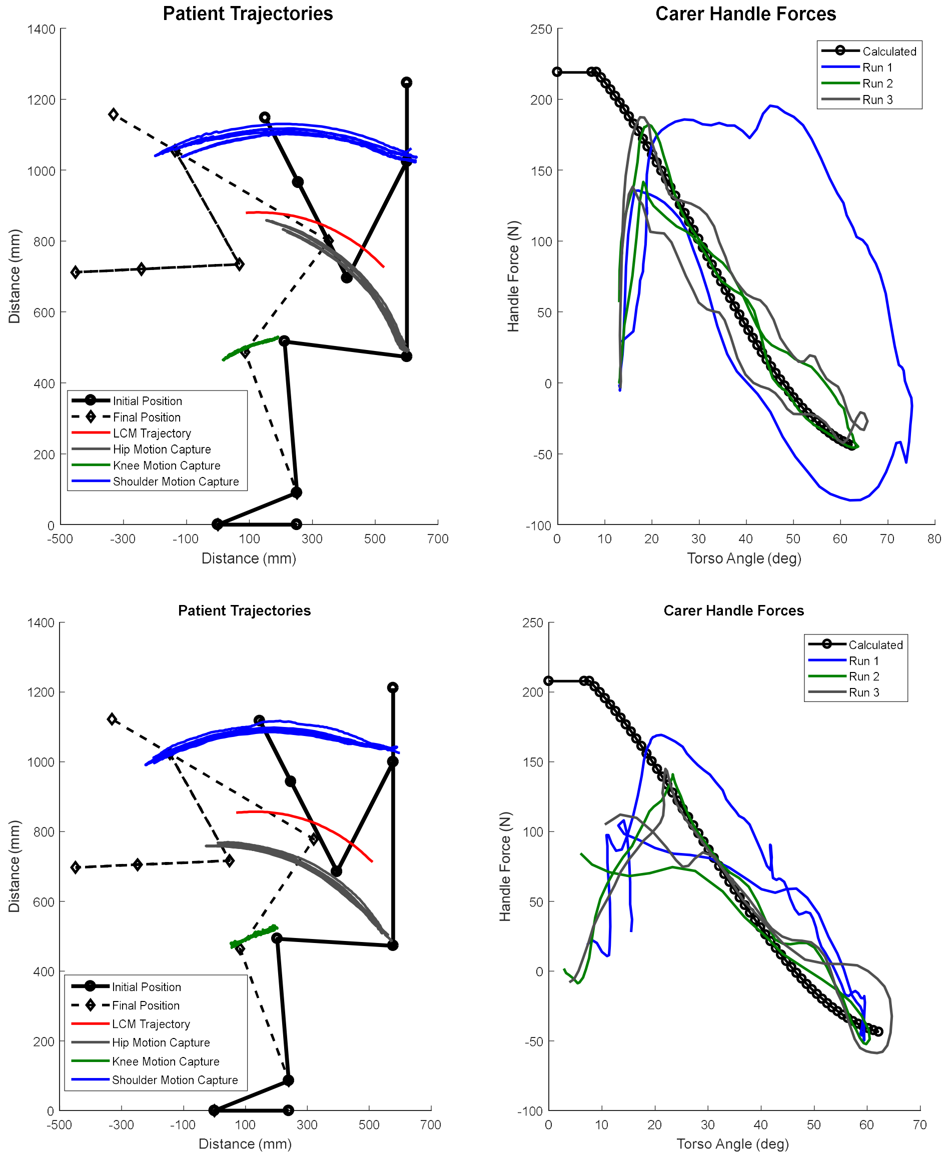




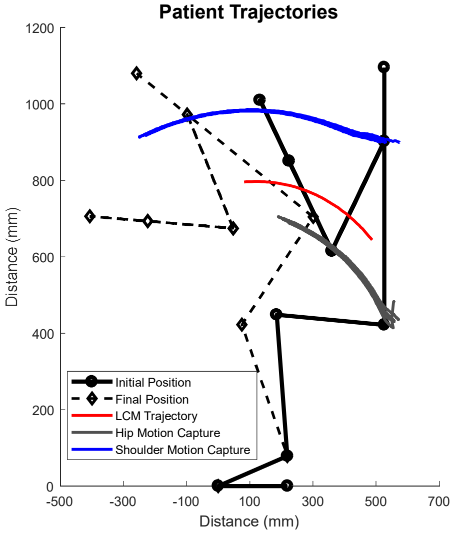
Patient F. At 490 Millimetre Transfer Surface

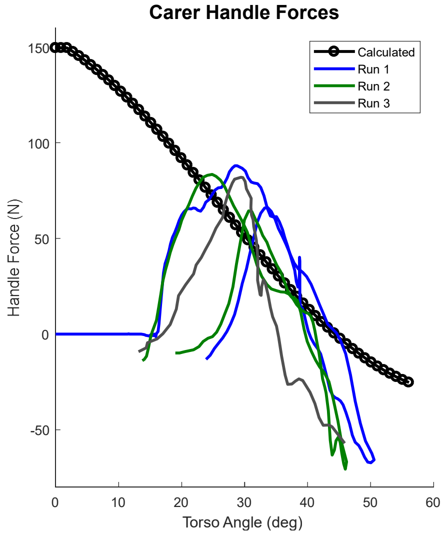


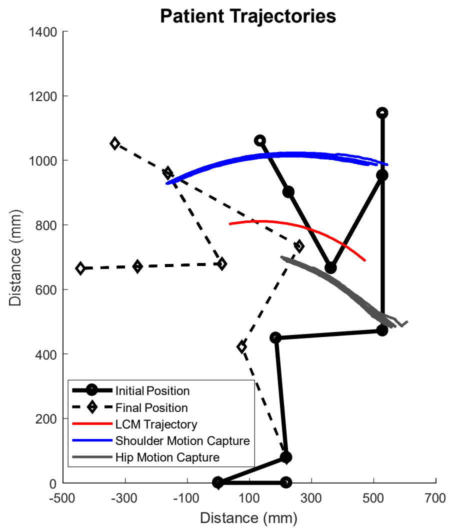
Patient G. At 460 and 490 Millimetre Transfer Surface

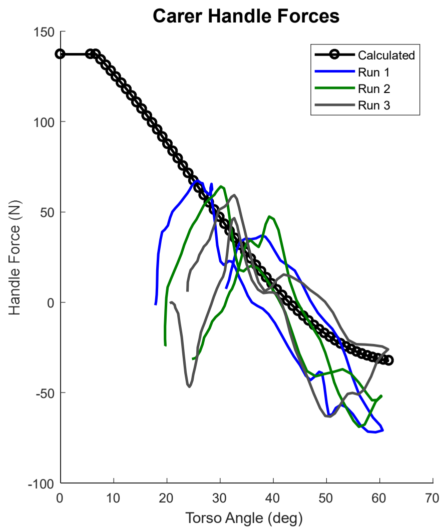


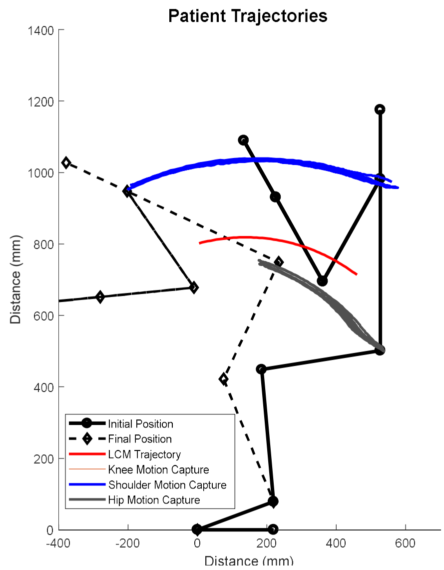
Patient H. At 460, 490 and 560 Millimetre Transfer Surface

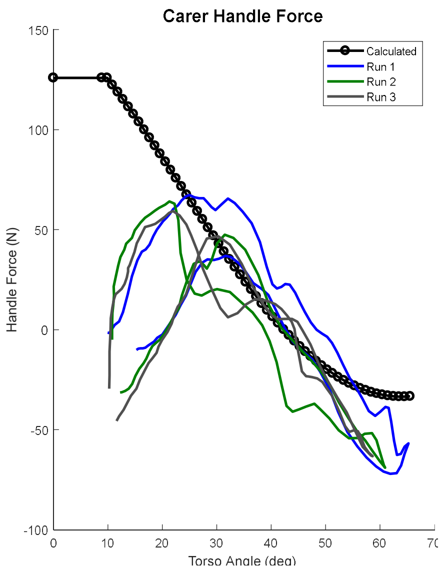




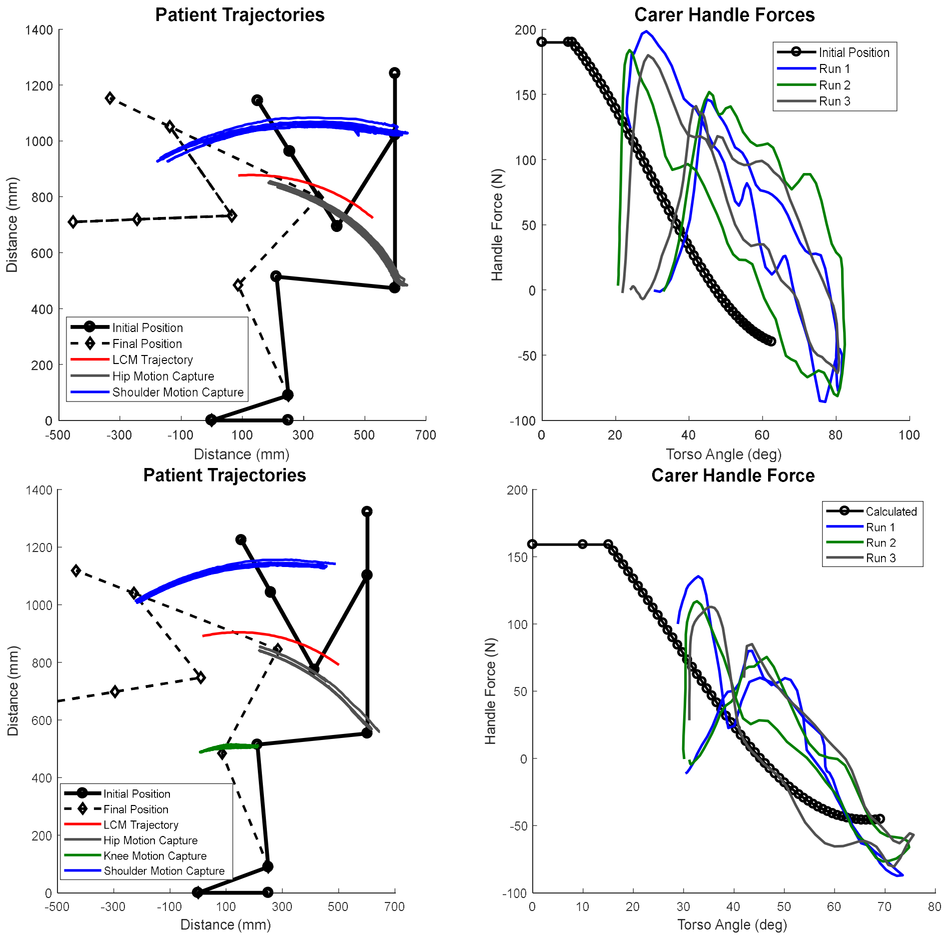




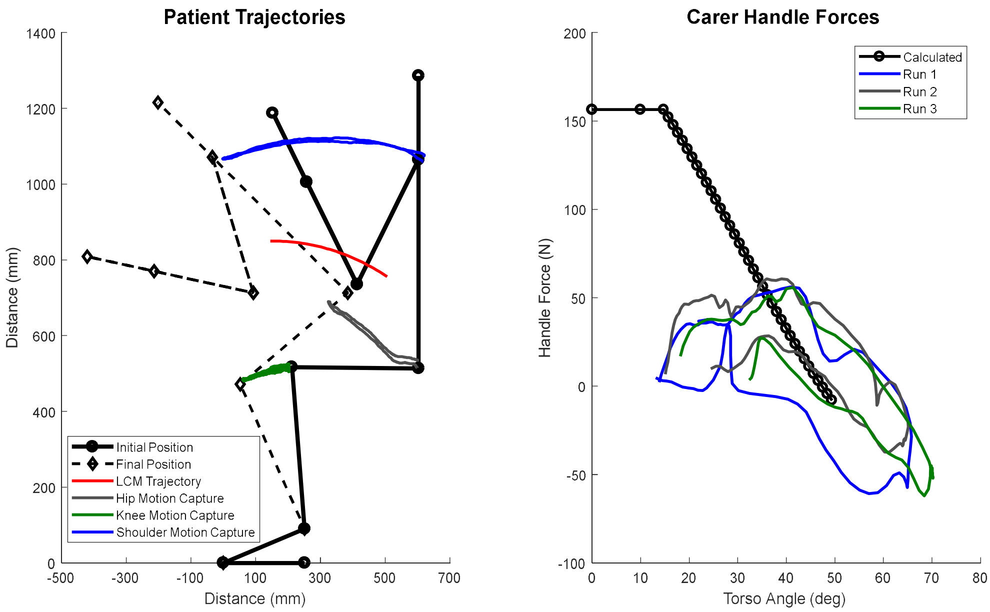




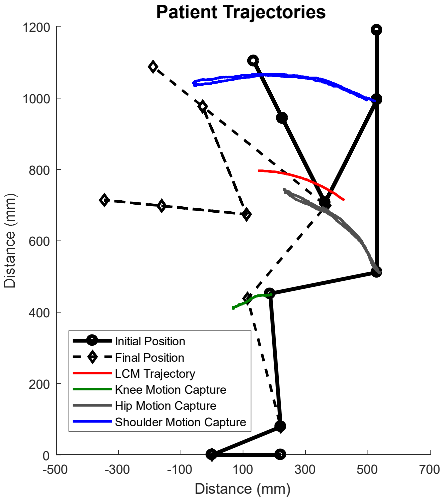
Patient I. At 460 and 560 Millimetre Transfer Surface

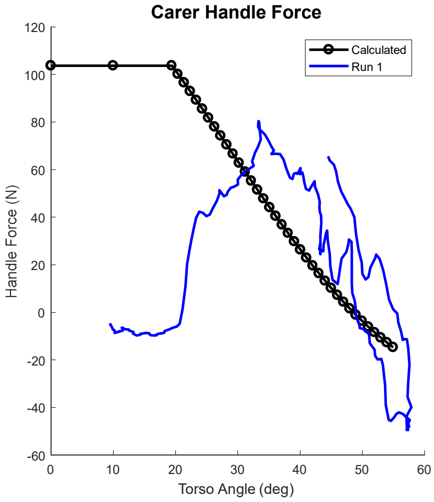


Patient J. At 490 Millimetre Transfer Surface

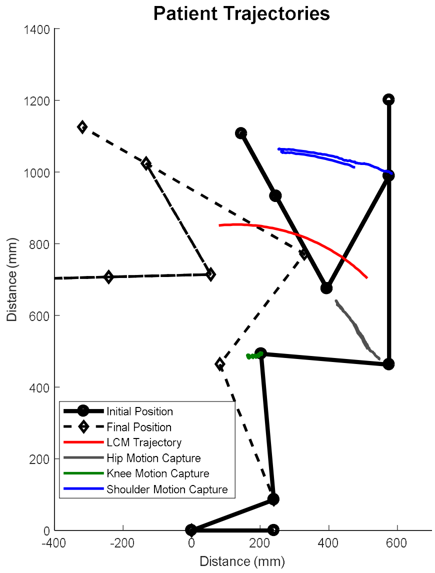


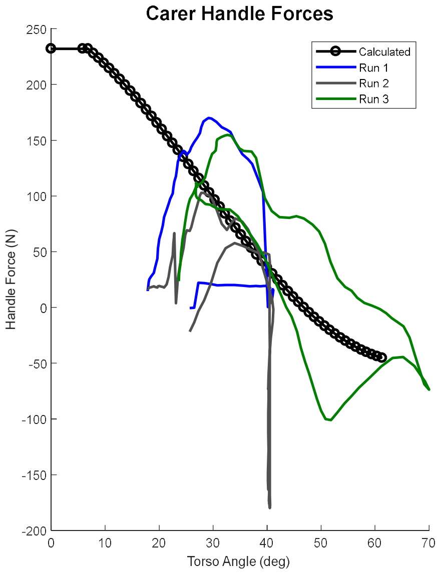
Patient K. At 490 Millimetre Transfer Surface





Patient L. At 490 Millimetre Transfer Surface





APPENDIX J THEORY SUMMARY

Figure 75 and Figure 76 summarise the theory developed to assess patient and lifter forces.

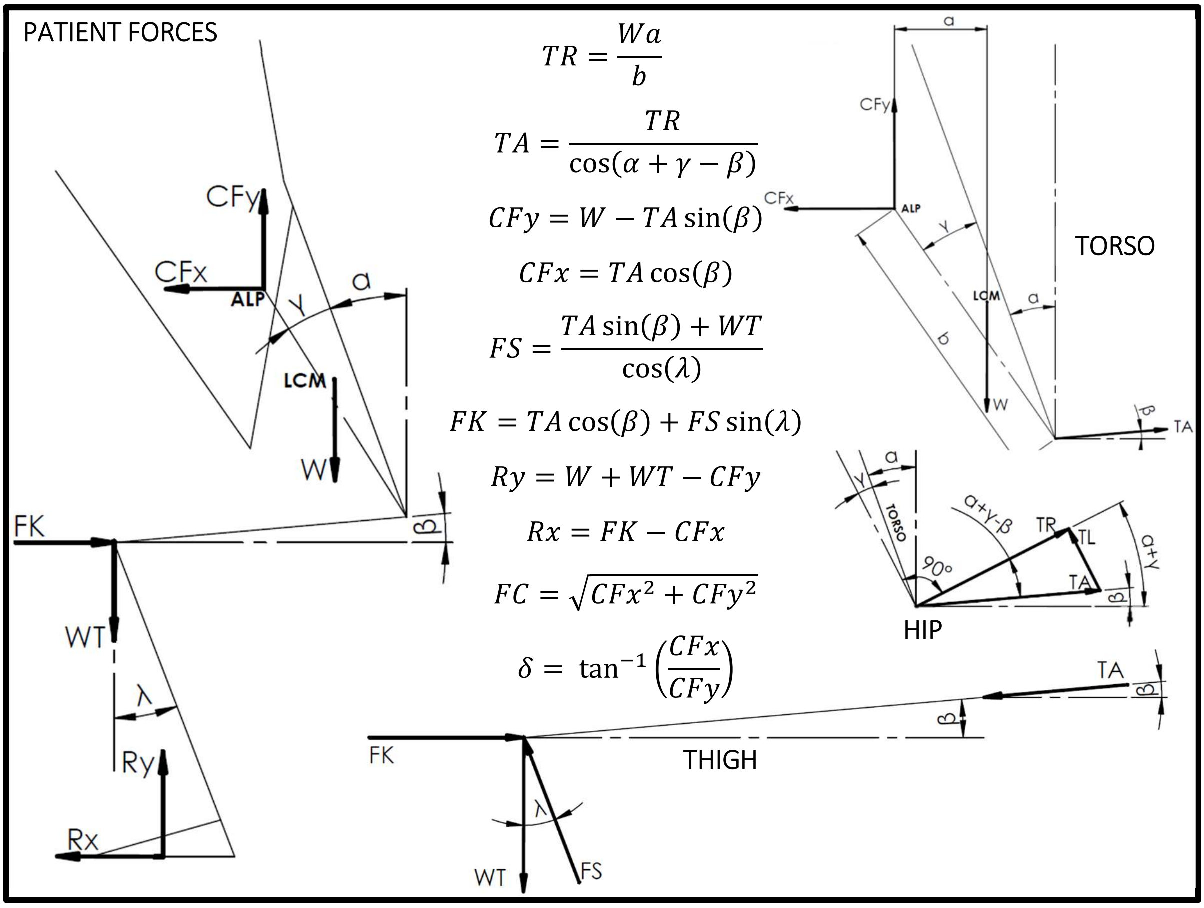


Figure 75 Summary of Patient Forces

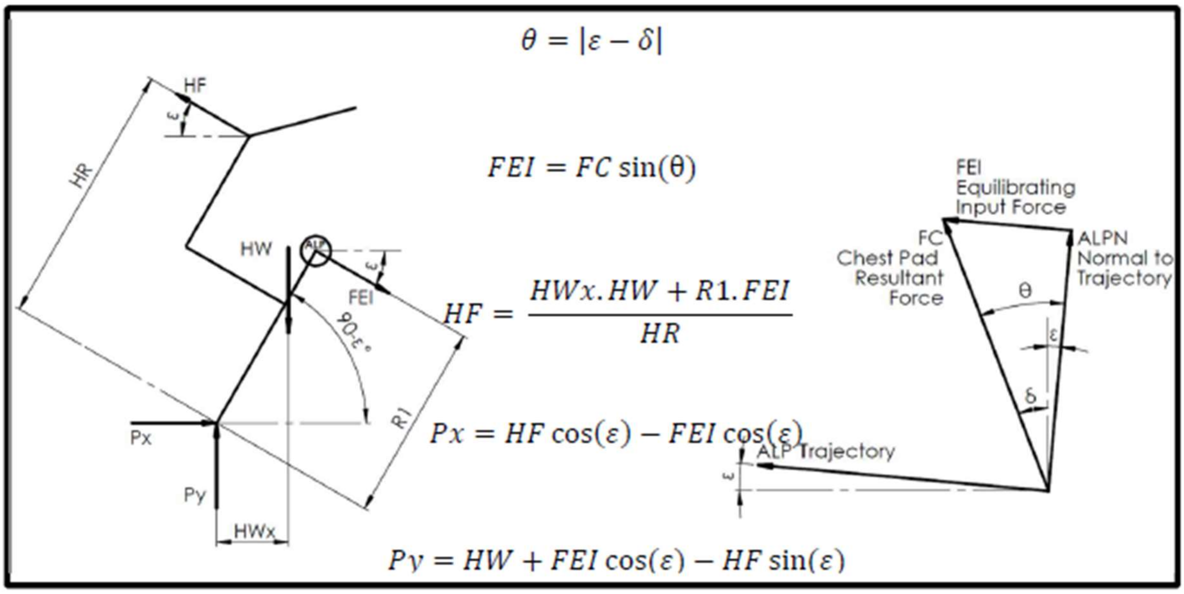
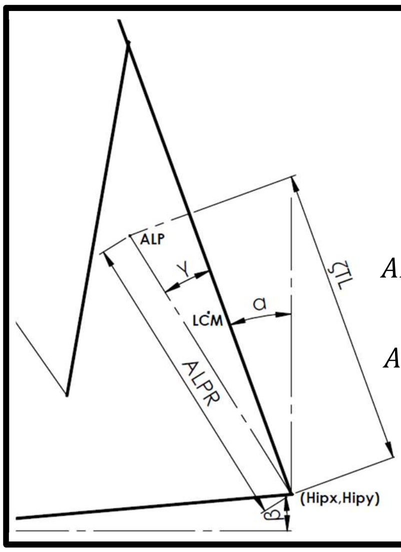


Figure 76 Summary of Lifter Forces

Figure 77 and Figure 78 summarise the theory developed to produce a zero force trajectory and how to approximate this with a single pivot lifter



$$\lim_{\varepsilon \to \delta} FEI = 0$$

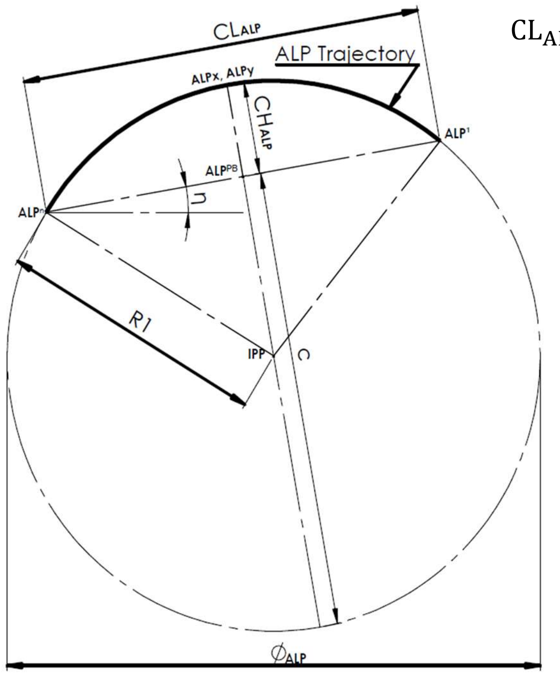
$$ALPR = \sqrt{\zeta {L_T}^2 (1 + \tan^2 \gamma)}$$

$$ALPx = Hip_x - ALPR \cos(90 - \alpha - \gamma)$$

$$ALPy = Hip_y + ALPR \sin(90 - \alpha - \gamma)$$

$$\varepsilon_n = \tan^{-1}\left(\frac{ALPy_n - ALPy_{n-1}}{ALPx_n - ALPx_{n-1}}\right)$$

Figure 77 Theory of ALP Placement and Angle to Generate a Zero Force Lift



$$\mathrm{CL_{ALP}} = \sqrt{(\mathrm{ALPx_n} - \mathrm{ALPx_1})^2 + (\mathrm{ALPy_n} - \mathrm{ALPy_1})^2}$$

$$\eta = \tan^{-1}\left(\frac{\mathrm{ALPx_n} - \mathrm{ALPx_1}}{\mathrm{ALPy_n} - \mathrm{ALPy_1}}\right)$$

$$\tan(\eta) = \frac{\mathrm{ALPx} - 0.5\mathrm{CLx_{ALP}}}{\mathrm{ALPy} - 0.5\mathrm{CLy_{ALP}}}$$

$$\mathrm{c} = \frac{\mathrm{CL_{ALP}}^2}{4\mathrm{CH_{ALP}}}$$

$$\varnothing_{\mathrm{ALP}} = \mathrm{CH_{ALP}} + \frac{\mathrm{CL_{ALP}}^2}{4\mathrm{CH_{ALP}}}$$

$$\mathrm{R1} = \frac{\mathrm{CH_{ALP}}}{2} + \frac{\mathrm{CL_{ALP}}^2}{8\mathrm{CH_{ALP}}}$$

$$\mathrm{IPPx} = \mathrm{ALPx} + \mathrm{R1}\sin(\eta)$$

$$\mathrm{IPPy} = \mathrm{ALPy} - \mathrm{R1}\cos(\eta)$$

Figure 78 Summary of Zero Force Approximation Technique for Single Pivot Lifter